STRESS and STRAIN in BONES

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STRESS and STRAIN in BONES

Their Relation to Fractures and Osteogenesis

By

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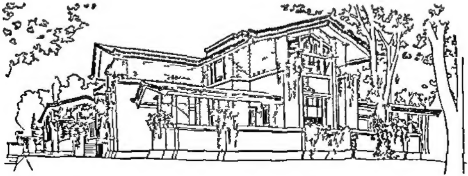

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Dedicated to
My Wife Harriet
A Constant Source of Inspiration

Preface

The present book is an attempt to present in one volume some of the more recent experimental work on stress and strain in bones and bone Most of the attention has been centered on investigations involving the use of actual bones or sections of bones, although analysis of stress and strain in plastic models of bones and in the intervertebral discs have also been included Many excellent but somewhat more theoretical and speculative studies, by means of diagrams, of the biomechanical behavior of bones under assumed conditions of loading have been omitted Throughout the discussion particular emphasis has been placed on the relation of stress strain phenomena to fractures and osteogenesis

In preparation of the manuscript the author is especially indebted to the following friends and colleagues for their many helpful suggestions and criticisms Professor Herbert R Lissner, Chairman of the Department of Engineering Mechanics Wayne University College of Engineering read the entire manuscript with particular attention to the parts dealing with engineering techniques and terminology Drs Ernest Gardner F Morin and Gabriel Lasker of the Department of Anatomy Wayne University also were kind enough to read all or parts of the manuscript

For the photographic illustrations the author is indebted to Mr Charles Pickard Department of Medical Illustration Wayne University, while the excellent diagrams were drawn by Mrs Ruth Rosenbaum Department of Anatomy Wayne University Special thanks are expressed to Miss Faith Brandt for her careful typing of the manuscript

The author also wishes to express his special appreciation to Dr Gordon H Scott Dean of the College of Medicine, Wayne University, for the active interest he has always shown in the author s own research on the biomechanical behavior and physical properties of bones These investigations have been made with the complete cooperation of Professors Herbert R Lissner and

Milton Lebow of the Department of Engineering Mechanics Wayne University

The author also wishes to express his appreciation to the publishers of the following journals for their permission to reproduce figures appearing originally in their publications *Acta Anatomica Acta Odontologica Scandinavica, American Journal of Anatomy American Journal of Physical Anthropology, American Journal of Surgery Anatomical Record Anatomische Nachrichten, Annals New York Academy of Science Archiv f Klinische Chirurgie, Bulletino delle Scienze Medicine (Bolona) Instructional Course Lectures* American Academy of Orthopedic Surgeons *Journal of Bone and Joint Surgery, Journal of Applied Physiology Journal of Neurosurgery, Librairie des Sciences, Bruxelles Radiology Surgery Gynecology and Obstetrics, Zentralblatt f Chirurgie*

The preparation of this book as well as most of the author s other research on the biomechanics of bone has been aided in part by research grants from the National Institutes of Health U S Public Health Service The author s investigations on the pelvis and vertebral column were supported by a research contract with the Aero Medical Laboratory Department of the Air Force

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Contents

	Page
Preface	vii
1 Introduction	3
2 Methods of Studying Stress and Strain in Bones	9
3 Mathematical Analysis of Stress and Strain in Bones	16
Summary	25
4 Stress Strain Studies with Models of Bones	26
Summary	33
5 Stress and Strain in Skull Deformation and Fracture	34
Dynamic Loading Studies	34
Static Loading Studies	49
Split line Studies	51
Summary	53
6 Stress Strain Distribution in Long Bones	54
Results With the Colophonium Method	54
Results With the Stresscoat Method	60
Results With Electric Strain Gages	71
Summary	72
7 The Magnitude of Stress and Strain in Long Bones	73
Measurements With Extensometers	73
Measurements With Stresscoat Lacquer	80
Summary	81
8 The Relation of Stress and Strain to Fracture of Long Bones	82
Summary	90
9 Stress and Strain in the Pelvis Mandible and Vertebral Column	92
The Pelvis	92
The Mandible	98
The Vertebral Column	101
Summary	110

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Detroit Michigan F GAYNOR EVANS, PH D

STRESS and STRAIN in BONES

10 The Role of Stress and Strain in Bone Architecture 111
Summary 128

11 The Effect of Stress on Bone Healing and Growth 130
Compressive Stress in Bone Healing 130
Compressive Stress in Bone Growth 140
Tensile Stress in Bone Growth 144
Summary 146

12 Stress and Strain During Embryonic Development 147
Effect of Stress in Embryonic Osteogenesis 147
Theories on the Influence of Stress in Osteogenesis 155
Summary 158

13 Factors Influencing the Breaking Strength of Bones 160
The Influence of Diet 160
The Influence of Hormones 168
The Effect of Disuse 169
The Influence of Healing 172
The Effect of Fluorosis 172
Summary 174

14 The Tensile, Compressive and Shearing Strength of Bone 176
Tensile Stress and Strain 177
Compressive Stress and Strain of Compact Bone 192
Compressive Strength of Spongy Bone 197
Shearing Strength of Bone 199
Summary 201

15 Torsion Bending and Fatigue Strength of Bone 203
Torsion Strength of Bone 203
Bending Strength 205
Fatigue Strength 225
Summary 226

References 229

Glossary 239

Index 241

1

Introduction

The earliest mention of the mechanical significance of bone form was made by Galileo (1638), but its strength and other physical properties were not extensively studied until the latter half of the nineteenth century. This was followed by a period of some 20 years before the subject was again investigated to any extent. Nowadays the great number of automobiles and the increasing popularity of air travel lend urgency to the question of the type and magnitude of the stresses and strains the human body can safely tolerate. Crash pads, windshields that fall out under impact, and better visibility for the driver are some of the safety features seriously considered by manufacturers and purchasers of automobiles.

Various branches of the armed services are also interested in the problem and are supporting basic research on the mechanical strength of the human body. The stresses and strains to which the pilot is subjected in jet fighter planes or in emergency escape from an airplane are especially important.

The rapidly increasing rate of accidents is being studied by special groups, notably the Crash Injury Research group affiliated with Cornell University Medical School, organized to collect data on automobile and airplane crashes. The data collected include the speed of the vehicle at the time of crashing, the position the injured person occupied in the vehicle, the direction the person faced, the medical evaluation of the injury, and the structural part of the vehicle believed to be responsible for the injury.

One of the most common injuries found in survivors of car or airplane crashes is fracture of one or more bones. For this reason the present monograph was prepared in an attempt to assemble in a single small volume some of the available information on stress and strain in bones, especially as related to the fracture mechanism. Most of the studies to be discussed are experimental

measured in kilograms or pounds. The magnitude of a couple is the product of one of its forces and the distance between them. Units for couples may be computed in pound inches, pound feet, or kilogram meters. For example, one inch pound of torsion (torque) is produced by a couple consisting of two one pound forces acting at a distance of one inch from one another. The *moment* of a force about a point is the product of the force times the distance from that point to the action line of the force. The moment of a couple about any point in the plane of the couple is simply the product of one of the forces and the distance between them. This is true no matter where the point of the moment is located.

Stress and strain are terms often erroneously used as synonyms although each has a distinct meaning in mechanics. *Stress* (Figure 1 b) is the intermolecular resistance within an object to the action of an outside force which is applied to it. An example of stress in the human body is the internal resistance produced within the leg bones as the result of the compressive force applied to them by the body weight when standing erect. Stress cannot be seen or measured directly. However, its magnitude can be computed by various formulas, depending upon whether the object is being subjected to tension, compression, shearing, torsion, or bending. In these formulas it is assumed that the object is composed of a homogeneous and isotropic material, which is not true of a bone. Consequently, the usual formulas require modification before being applicable to a study of stress in bones.

Strain may be tensile, compressive, or shearing, the latter being a change in angle instead of a change in linear dimensions. *Total strain* (Figure 1 c) is the change in the linear dimensions of an object as the result of the application of a force. For example, if an object were one inch long before stretching and three inches long after stretching, it would have a total strain of two inches. *Unit strain* is the ratio between the change in dimensions and the original length of the object. The magnitude of unit strain is obtained by dividing the total deformation of the object by its original length. The significance of unit strain may be emphasized by assigning to it the unit, inches per inch, which indicates that it represents the change in length per unit of

in nature and involve the use of actual bones or models of bones Many excellent but more theoretical studies in which the forces presumably acting upon bones have been investigated and analyzed by means of diagrams are omitted

The study of stress strain phenomena in bones is a problem in biomechanics and its analysis requires the use of many of the techniques employed by engineers to describe similar phenomena in structural materials Therefore a brief review of some of the basic principles of mechanics may be helpful

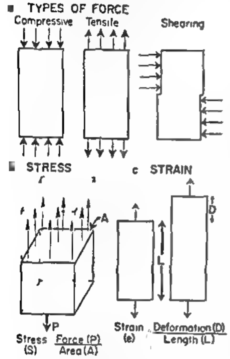

Figure 1 (From Evans *Ann New York Acad Sc* 63 1955)

The majority of the bones of *the body especially those of the* limbs serve as levers upon which the muscles act to produce movement During movement as well *as at rest the bones are subjected* to a variety of forces which are primarily the result of muscle action and body weight Consequently it is appropriate to begin with a discussion of forces

A *force* is most simply defined as a push or a pull There are three kinds of force (Figure 1 a)—tensile compressive and shearing *A tensile force tends to pull an object* apart *or lengthen it* while a *compressive* force has a tendency to push an object together or shorten it *A shearing force tends to make one part of an* object slide over an immediately adjacent part

The various kinds of force can act separately or in combination with one another Two parallel oppositely directed forces of equal magnitude but with different lines of action constitute a *couple* When a couple is applied to an object perpendicular to its long axis it produces a *torsion* or twisting causing one part of the object to be rotated with respect to the remainder

The magnitude of tensile compressive and shearing force is

measured in kilograms or pounds The magnitude of a couple is the product of one of its forces and the distance between them Units for couples may be computed in pound inches, pound feet or kilogram meters For example one inch pound of torsion (torque) is produced by a couple consisting of two one pound forces acting at a distance of one inch from one another The *moment* of a force about a point is the product of the force times the distance from that point to the action line of the force The moment of a couple about any point in the plane of the couple is simply the product of one of the forces and the distance between them This is true no matter where the point of the moment is located

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length in the object being studied The magnitude of the strain produced in an object can be measured by various types of extensometers and strain gages If a strain is sufficiently large it can be seen but stress is always a derived quantity

Some of the principles of stress and strain can be demonstrated by the behavior of a column subjected to various types of loading This analogy is especially pertinent with reference to the long bones which, in the living body are constantly subjected to bending and twisting by the actions of muscles and the weight of the body

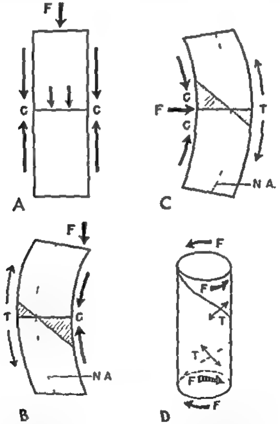

Figure 2 (From Evans Instr Course Lect *Am Acad Orthop Surg* 9 1952) The heavy arrow indicates the point of application of the force or load the light arrows the direction of the strain C = compressive strain F = force or load T = tensile strain N A = neutral axis or plane

A concentrically loaded column (Figure 2 a) is subjected to compressive stress and strain throughout the area of its cross section However if a column is eccentrically loaded (Figure 2 b) it is bent in addition to being compressed As a result of the bending tensile stresses and strains are created on the convex side of the bent column while compressive stresses and strains arise on the concave side of the column The magnitude of the stresses and strains is greatest at the surface of the column and gradually decreases to zero toward the center of the column The plane at which the magnitude of the stress is zero is called the neutral plane The more eccentric the loading the greater will be the bending of the column and the magnitude of the resulting stresses and strains

Loading a column from the side, perpendicular to its long axis (Figure 2 c) produces similar bending with consequent tensile and compressive strains and stresses on the convex and concave aspects respectively of the bent column When couples are applied to the ends of a column (Figure 2 d) the ends are twisted or rotated in opposite directions to one another This twisting produces tensile strains and stresses which spiral around the column at a 45 angle to its long axis

When the stresses produced in a material are plotted against the corresponding strains a *stress strain* curve is obtained The slope of this curve represents the *modulus of elasticity* of the material or the factor of proportionality between stress and strain The modulus of elasticity is a measure of the stiffness of a material and its magnitude is obtained by dividing the unit stress by the unit strain when the stress strain curve is a straight line The actual value for the modulus is commonly expressed in kilograms per square centimeter or pounds per square inch

Elasticity is the physical property of a material which permits it to return to its original form or shape after it has been deformed by a load or force Few materials not even a rubber band are completely elastic So practically all materials have an elastic limit or point of maximum stress beyond which the material will not return to its original shape

Another important concept in mechanics is that of *energy* which may be defined as the capacity to do work In most in

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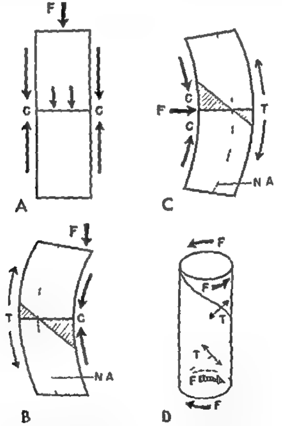

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2

Methods of Studying Stress and Strain in Bones

STRESS AND STRAIN in bones may be studied by the following methods (1) studying sections or pieces of bone, (2) recording stress strain phenomena in models of bones and (3) analyzing stress strain phenomena in intact bones

(1) The first method is historically the oldest and has been used most extensively Most investigators sectioned different bones in various planes and studied the sections with the hope of determining the functional significance, in terms of stress and strain of the orientation of the bony trabeculae Thus von Meyer (1867), Roux (1885), Wolff (1870 1892), Janssen (1920), and Carey (1929), to mention a few, studied the trabecular orientation in sections of bones with respect to their presumed function of resisting tension or pressure Elaborate and detailed mathematical analyses of entire cross sections of long bones under assumed conditions of loading were made by Koch (1917), Grunewald (1920) and Marique (1945)

Other investigators, e g Wertheim (1847) Rauber (1876) Hulsen (1896) Evans and Lebow (1951 1952) and Dempster and Liddicoat (1952) cut pieces of bone to standardized size and determined their various physical properties and biomechanical behavior under various types of stress and strain The techniques used were those employed by engineers in analyzing the strength of materials

(2) The study of stress and strain in bones by analysis of similar phenomena in models has also been used The first attempt was Culmann s famous trajectorial diagram published by von Meyer (1867) of the lines of maximum internal stress in a Fairbairn crane which he assumed resembled the femur in shape and in loading Culmann s analysis forms the basis for the tra-

vestigations of the strength of bones the energy was applied by dropping an object upon the bone being tested Under these conditions the magnitude of the energy applied to the bone can be computed by multiplying the weight of the falling object by the distance through which it travelled before striking the bone The amount of energy absorbed by the bone (per unit volume) up to the point at which it failed may be determined by measuring the area beneath the stress strain curve for the bone

In ascertaining the load a body can support before failure the speed of loading is an important consideration A greater load can be supported without failure if it is slowly (statically) applied than if it is suddenly (dynamically) applied However the magnitude or the extent of the strain or deformation produced in an object by a small load suddenly applied may be equivalent to that arising from a much greater load slowly applied

Under dynamic conditions i e when the load is suddenly applied energy absorbed is the fundamental criterion rather than force The latter can only be identified under static conditions when its magnitude, direction and line of action are known Thus it is necessary to speak of the energy of a blow rather than the force of a blow Energy and force cannot be directly related to one another as they are not the same and have different units of measurement Thus, the magnitude of energy is measured in kilogram meters or inch pounds while force is measured in kilograms or pounds When a load is statically or slowly applied the term force can be more appropriately used because its three distinguishing characteristics can be determined

The terms defined and the principles discussed briefly above are those most frequently involved in studies on the biomechanics of bones Their relation to specific investigations of stress strain phenomena in bones will appear in subsequent chapters

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studied Thus a three dimensional concept of the object can be obtained and by appropriate methods the stresses can be calibrated and accurate mathematical analyses made of the actual internal stresses in the original object Studies by means of photoelasticity of the stress and strain in models of bones have been made by Hallermann (1934), Milch (1940), and Pauwels (1948 1951)

(3) Other industrial techniques employed for stress strain analysis of engineering structures have been used for studying similar phenomena in intact bones The over all strain distribution in an object can be studied by coating it with a strain sensitive lacquer while the actual magnitude of the strain can be measured with extensometers and strain gages

The first investigator to use a strain sensitive lacquer for studying stress and strain in bones was Kuntscher (1934 1935 a and b) who used a method developed by Dietrich and Lehr of the Maybach Company for determining points of weakness and failure in machine parts The object to be tested was coated with melted colophonium which after drying cracked in response to tensile strain or deformation in the underlying material The resulting cracks were transverse to the direction of the tensile strain and indicated sites of highest tension where failure or fracture would occur under sufficient load Because of its low tensile strength the colophonium cracked at a stress or load far less than the elastic limit of the object being tested

Kuntscher applied the method to a study of the deformations or strains occurring in the long bones of the human body under different conditions of loading and orientation He found that if the colophonium were applied while still boiling it easily blistered and could not be used Therefore it was applied to the bone after being evenly heated to slightly less than 40 C The best results were obtained when the layer of colophonium applied to the bone was approximately 1 mm in thickness The sensitivity of the colophonium was approximately 0 001 inches per inch i e every time the underlying material was stretched 0 001 inches a crack would appear in the overlying colophonium

If Kuntscher wished to examine the bone for tensile strain he removed it from the melted colophonium after the bubbles

jectorial theory of bone architecture, although as will be discussed later it has been severely criticized on many points. Shortly after Culmann, Roux made a rubber model of an ankylosed knee joint which he coated with paraffin. He then loaded the model of the joint in the way he assumed had occurred in the living individual and, from the cracks produced in the paraffin, drew trajectorial diagrams which he believed indicated the functional significance of the orientation of the trabeculae in the actual specimen. A more detailed discussion of this study, as well as the criticisms advanced against it, will be given later.

Within the last few decades stress strain phenomena in bones has been studied by the use of an engineering technique based on photoelasticity. The method is used in industry for stress strain analysis of machine parts. It consists of loading a plastic model of the object being analyzed and studying by means of polarized light the stress patterns produced in the model. The method depends on the fact that phenylformaldehyde resins are birefringent under stress and behave somewhat like temporary crystals whose molecular structure is oriented along the lines of stress.

When the plastic models are examined under polarized light the stress lines become evident. With white light the stress patterns exhibit all the colors of the spectrum but in engineering practice monochromatic light is generally used. The stress patterns are not changed quantitatively or qualitatively by monochromatic light and in photographs the patterns appear as alternating bands of light and dark.

In analyzing the significance of photoelastic patterns it should be pointed out that the alternating dark bands obtained in the plastic model are proportional to the magnitude of the shearing stresses within the model. However, at the margin or boundary of the model the bands are proportional to the tensile or compressive stress because at any free margin the magnitude of the shearing stress is equal to half the tensile or compressive stress present. The alternating dark bands do not show the direction of the maximum tensile and compressive stress in the model which, however, can be determined from an isoclinic model.

If the model is composed of a thermoplastic resin it can be cooled under stress, sectioned, and its internal stress pattern

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If the model is composed of a thermoplastic resin it can be cooled under stress sectioned and its internal stress pattern

with the stresscoat lacquer. If a dark undercoating material is used the cracks appear as white upon a dark background. The cracks in the stresscoat lacquer may also be visualized by painting the area with a red dye etchant, which is allowed to remain on the object a short time before being removed. However, the dye changes the sensitivity of the lacquer and the object must be recoated before another test can be conducted. For this reason the dye is less desirable than the statiflux powder.

At the time of a test the sensitivity of the lacquer on the calibration strips is determined in inches per inch, which is also a measure of the tensile strain necessary to cause the lacquer to crack. For example, if the sensitivity of the lacquer is 0.0005 inches/inch it means that when an inch of the underlying material has stretched as much as 5/10 000 of an inch (0.0005 inches) the overlying material will crack. Thus every crack indicates a stretching or tensile strain of the underlying material by at least 0.0005 inches. The sensitivity of the lacquer is markedly influenced by the temperature and humidity conditions at the time of the test.

The stresscoat lacquer has a higher sensitivity than does the colophonium and since is does not need to be heated is much easier to use as it is applied to the test object with a spraygun. Consequently the lacquer is well impregnated with minute air bubbles so that a crack occurring in it is constantly interrupted by new air bubbles and must, therefore, continue to be propagated only by the tensile strain in the underlying material. Although cracks in the colophonium also arose from tensile strain in the underlying material their continued propagation did not depend upon the strain conditions in the material. The end of every crack in the colophonium was an area of high stress concentration, and the cracks continued to be propagated in their initial direction without regard to the actual stress conditions in the underlying material. This was the result of the homogeneity of the colophonium as there were few air bubbles which could interrupt the propagation of the cracks.

The stresscoat method was first applied to a study of stress and strain in bones by Gurdjian and Lissner who with their co-workers have used it in an extensive series of studies on skull

had disappeared, and allowed it to cool If compressive strain were to be studied the bone was placed in a testing machine and then coated with the colophonium which was permitted to cool In tensile testing the colophonium cracks appeared as the load was gradually increased in compressive testing they appeared as the load was slowly decreased The area of the bone in which the cracks first appeared was the site of highest tensile strain In tensile testing the strain was created by the pull or tension exerted upon the entire system In compressive testing the cracks appeared in the areas formerly under compression from the ten sion or stretching of the bone as the load was gradually removed The method gave a fairly good over all picture of the strain dis tribution in a bone subjected to tensile or compressive loading

Recently the distribution of tensile strain in whole bones has been restudied by the Stresscoat method developed by deForest and Ellis (1940) The principles of the method are almost iden tical with the colophonium technique but stresscoat is easier to use and more sensitive than colophonium The test object is sprayed with an aluminum or dark undercoating which is allowed to dry before stresscoat, a brittle resinous lacquer, is applied The stresscoat is evenly applied by a spray gun and permitted to dry for 24 hours before the object is tested Steel calibration strips are coated at the same time as the test object The grade or number of stresscoat lacquer used depends upon the temperature and humidity conditions certain grades being better for winter and others for summer Like colophonium stresscoat cracks in re sponse to tensile strain in the underlying material the cracks lying transverse to the direction of the strain and first appearing in the area of highest tensile strain

Just before testing the stresscoated object is coated with a liquid sensitizing material which is wiped off as soon as the test is completed The object is then sprayed with a white powder called Statiflux which because of its electrostatic charge col lects along the cracks in the stresscoat lacquer and makes them visible

The statiflux powder does not change the sensitivity of the lacquer and if no cracks are visible another test with greater load or energy can be made without having to recoat the object

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deformations and fractures (1945, 1946 1947, 1950, 1953) It has been similarly employed by Evans, Lissner, and Pedersen (1948, 1949 1951 1952, 1953) in studies on femoral deformations and fractures Evans Hayes, and Powers (1953), using the stresscoat technique have computed the approximate magnitude of the tensile stress produced by dynamic transverse loading of the human femur

The great advantage of strain sensitive lacquers is that they enable one to visualize the over all strain distribution produced in an object under controlled conditions of loading and orientation Furthermore a photograph can be taken of the pattern and then the lacquer can be removed a new coat applied and additional tests made upon the same object Thus a comparison can be made of the type and distribution of strain produced in the same object under various conditions of loading and orientation In addition, if one wishes to take very accurate measurements of the magnitude of the strain produced in an object under given experimental conditions the strain pattern produced in the lacquer indicates the site where extensometers or strain gages should be placed

The magnitude of the tensile and compressive strains produced in intact bones under various conditions of loading and of orientation has also been measured by different types of extensometers Kuntscher (1936) used Okhuizein extensometers for measuring the tensile and compressive strains produced in the neck of the femur under various loads The extensometers had a gage length of 10 to 20 mm and an accuracy of 0 1% Marique (1945) also made a study of the femoral neck and shaft using Huggenberger extensometers of an accuracy similar to those employed by Kuntscher

Recently the tensile and compressive strains occurring in bones of living dogs have been recorded by wire resistance strain gages connected with cathode ray oscilloscopes Gurdjian and Lissner (1944) used such gages in studying the strains produced in the skull of the living dog under loads of various magnitudes This method has also been used by Evans (1953) in collaboration with Coolbaugh and Lebow in recording the strains produced in the tibia of the living walking dog By proper calibration

of the gages with the oscilloscopes the actual magnitude of the various types of strains can be computed

Wire resistance strain gages are based upon the principle that when a wire is stretched or shortened its electrical resistance to the passage of a current changes in proportion to a change in the length of the wire These changes in resistance can be determined by electrical means and are the basis for measuring the changes in the length of the wire and therefore, similar changes occurring in the material to which the electric strain gage is cemented

The gages are generally made from an alloy of nickel copper (45% nickel and 55% copper) wire 0 001 inches in diameter For each gage the ratio of the unit change of resistance to the unit change of length of the wire is constant This ratio is called the gage factor which is 2 1 for most gages made in this country However the factor varies for different wires used in making the gage Thus a strain gage having a gage factor of 2 1 would have an increase of resistance of 0 21% when its length is changed by 0 1% in stretching or shortening When the wire is stretched (tensile strain) its length increases while its diameter decreases, in proportion to a ratio (Poisson s) for the particular metal used in making the wire By embedding the gage in a plastic material to prevent it from buckling a strain gage of this type can also be used for measuring compression Because of their extreme accuracy 0 000001 inches under optimal conditions in measuring minute changes in the linear dimensions of the surface to which they are attached gages of this type have been extensively used in industry since 1938

In subsequent chapters the results obtained in studying stress and strain in bones by each of the above described methods will be discussed fully

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3

Mathematical Analysis of Stress and Strain in Bones

ALL THE METHODS of stress strain analysis discussed in Chapter Two have been applied to human long bones, especially the femur. The results obtained by mathematical analysis of cross sections of bones will be discussed first, because of the historic role such analysis has played in the development of various theories of the functional architecture of spongy bone.

Mathematical analysis of sections of bones was used by Koch (1917), Grunewald (1920), and Marique (1945) in studying stress and strain in the long bones. Some of the difficulties encountered, as pointed out by Marique, are (1) bending of the bone takes place in more than one plane, and (2) the shape of the transverse section as well as the ratio between the amount of compact and spongy bone constantly changes along the length of the bone. Consequently the moment of inertia and the center of gravity change from one section to another. In addition, because of the asymmetrical shape of the cross section, the principal axes (of inertia) about which bending occurs and the exact location of the neutral axis also change along the bone. Therefore all these characteristics must be determined separately for each section in order to compute the stress and strain of the section. Although Marique's remarks referred to the femur, they apply equally well to mathematical analysis of sections of any bone. The ways in which different investigators have attempted to solve these difficulties will now be discussed.

The most thorough study of stress and strain in a bone by mathematical analysis of cross sections is Koch's classic paper (1917) on the *Laws of Bone Architecture*. Koch was particularly qualified for such a study because he was a professional civil engineer before becoming a Doctor of Medicine.

Koch first made a preliminary study of some 25 femurs from dissecting room cadavers but did not attempt a detailed mathematical analysis, since no data were available about the individuals whose bones were studied Finally he obtained the femurs of a healthy 200 pound, six foot tall Negro man about 35 years of age, who had been accidentally killed Comparison of these femurs with those of 30 dissecting room cadavers showed them to be typically normal bones The right femur was cut, at ¼ inch intervals into 75 cross sections the majority of which were subjected to detailed quantitative mathematical analysis The left femur was cut into serial frontal sections

Koch in accord with American engineering practice, gave his values for stress and strain in terms of pounds per square inch (lbs/in^2) while those of Grunewald and Marique are given in kilograms per square centimeter (kg/cm^2) Pounds per square inch may be converted into kilograms per square centimeter by multiplying by 0 0703 Kilograms per square centimeter can be changed into pounds per square inch by multiplying by 14 22

The computations of stress were based upon an assumed vertically applied load of 100 pounds in standing 160 pounds in walking and 320 pounds in running Koch believed the mechanical effects of body weight

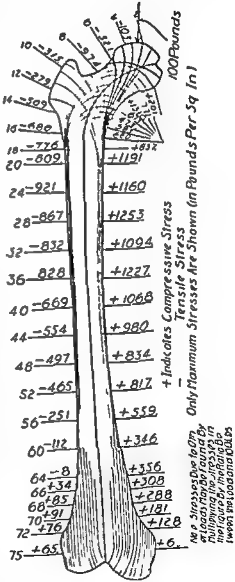

Figure 3 Computed stresses in a femur under vertical loading (From the original Figure 18 Koch *Am J Anat* 21 243 1917)

were far more important in determining bone architecture than was muscle action and ignored the latter influence. Figure 3 gives the values obtained, according to section, for the maximum tensile (−) and compressive (+) stress produced by a 100 pound load. The magnitude of the stresses should be multiplied by 0.6 for standing "at attention," by 1.6 for walking, and by 3.2 for running.

The maximum unit tensile stress (974 lbs/in^2) for the entire bone was found in the middle of the femoral neck (section eight). That for the shaft (921 lbs/in^2) occurred in section 24 near the junction of the proximal and middle thirds of the shaft. The areas of the bone subjected to tensile stress were the superior aspect of the neck and the lateral aspect of the shaft as far as section 64, distal to which the region was under compressive stress.

The entire medial aspect of the femur, including the inferior aspect of the neck, was subjected to compressive stress, the magnitude of which in each section exceeded that for tension. Section eight had the highest compressive stress (1310 lbs/in^2) for the entire bone. The highest compressive stress in the shaft (1253 lbs/in^2) was in section 28.

The magnitude of the assumed load did not change the level of the highest tensile and compressive stress, although the actual value for each type of stress was increased. Thus, under an assumed load of 160 lbs the tensile and compressive stress in section eight increased to 1560 lbs/in^2 and 2098 lbs/in^2, respectively. When the assumed load was 360 lbs the stress rose to 3117 lbs/in^2 and 4192 lbs/in^2 for tension and compression, respectively. The shearing stress in section eight amounted to 231 lbs/in^2 under a load of 100 pounds, 369 lbs/in^2 under a load of 160 pounds, and 400 lbs/in^2 with a load of 360 pounds.

In Grunewald's study (1920) the stresses and strains in the femur and tibia were computed on the basis of a vertically applied load of 30 kg, representing the body weight when standing in a comfortable position. For purposes of analysis the femur was cut into 11 cross sections: three in the neck, one apparently between the neck and the shaft, and seven in the shaft. The sections from the neck were designated *a*, *b*, and *c*. Section

a was nearest the head *b* in the middle of the neck and *c* next to the junction of the neck with the shaft The magnitude of the tensile compressive and shearing stresses was computed for the sections from the neck and the first six sections from the shaft

According to his calculations the maximum tensile (54 8 kg/cm²) and compressive (64 7 kg/cm²) stress in the neck were in section c which being a danger area, was reinforced by thickening of the compact bone on the inferior aspect of the neck Section *b*, the site of most femoral neck fractures, was weakly constructed and had no such thickening of compact bone The tensile stress on the convex (superior) aspect of the neck in sections *b* and *c* was 29 9 kg/cm² and 54 9 kg/cm² respectively The compressive stress on the concave (inferior) aspect of both sections was 54 4 kg/cm Section *a* had a shearing stress of 22 5 kg/cm² and a compressive stress of 21 kg/cm² According to Grunewald, these latter values were the same for the entire femoral neck because the direction of the neck and of the load remained constant In contrast to Koch s results the highest tensile (63 4 kg/cm²) and compressive (78 8 kg/cm²) stresses of the entire bone were found in the first section of the shaft

In analyzing the tibia Grunewald considered the weight line as falling in the neighborhood of the posterior border of the bone about in the center of the frontal plane The neutral axis was nearer the anterior border of the bone The load was assumed to be a body weight of 30 kg For purposes of mathematical analysis the bone was divided into a proximal and a distal half Under these conditions Grunewald found that the maximum tensile (7 0 kg/cm²) and compressive (47 7 kg/cm) stress occurred in the distal half of the bone The corresponding values for the proximal half of the tibia were 5 9 kg/cm² and 26 kg/cm², respectively Thus under the same load the magnitude of the tensile and compressive stresses developed in the tibia was considerably less than that in the femur

Marique s study (1945) of stress and strain in the femur was based on 16 cross sections taken from a left femur No data were given about the individual from whom the specimen was obtained although the right femur (Femur I) had previously been

used for measuring with Huggenberger extensometers, the deformations produced by static vertical loading.

The first of the 16 sections was through the approximate middle of the neck and the remaining ones, starting at the inferior border of the lesser trochanter, were sawed from the shaft at intervals of 2 cm. Roentgenograms were taken of each section. Because the roentgenograms were always slightly distorted by the x rays, which struck the object obliquely, the mathematical analyses were made on photographs which had been enlarged four times. Sections one, three, eight, and 12 were then analyzed in detail with respect to such factors as the moment of inertia, the axes of inertia, the center of gravity, the area density, and magnitude of the various types of stresses in the sections. As a control a transverse section from the neck of another left femur was studied. Presumably the load applied to the head of the femur was 100 kg.

The magnitude (kg/cm^2) of the various stresses in the femur, as computed by Marique, is summarized below.

Stress	*Neck Sect 1 Femur I*	*Neck Sect 1 Femur IV*	*Shaft Sect 3 Less Troch*	*Shaft Sect 8 Mid Shaft*	*Sect 12 Epicondylar Region*
Tensile (Max.)	117.14	103.84	105.82	68.20	11.77
Compression (Max.)	148.81	139.32	148.72	103.39	39.6

Marique's figures show that the neck of the femur was the site of highest tensile stress in the entire bone. The highest compressive stress occurred in the neck and at the level of the lesser trochanter. The magnitude of the compressive stress exceeded that for tension in each section, although both types of stress decreased distally along the shaft of the bone.

In comparing the results obtained by Koch, Grunewald, and Marique, one must consider the way in which each investigator tried to overcome the difficulties inherent in analyzing cross sections of long bones. The chief problem was computing the moment of inertia for sections having both spongy and compact bone. The moment of inertia in turn depends upon the method of determining the principle axes of inertia.

The moment of inertia of a section must be known, because its value is one of the factors used in the formula for calculating

the magnitude of the stress in the section The moment of inertia is in turn related to the principle axes of the section and the ratio between the amount of compact and of spongy bone in the section In cross sections composed mostly of compact bone, e g those of the shaft, there is little difficulty However, in sections through the neck of the bone where there is a large amount of spongiosa, computing the moment of inertia is more difficult because the various physical properties of spongy bone, especially its modulus of elasticity, are not as well known as those for compact bone

The moment of inertia is a mathematical expression used in engineering to represent a partial measure of the strength of a structural member Its use is dependent upon the fact that the design of the parts of an engineering structure is based upon the cross section area of the part and not upon the part as a solid body The moment of inertia is determined for an area about an axis (of inertia) around which bending occurs This may be done by dividing the area about the axis into a large number of small areas, the moment of inertia for the entire area being equivalent to the sums of the products obtained by multiplying each of the small areas by the square of its distance from the axis

Koch identified the axes of inertia by visual inspection and Grunewald by analogy with regular geometric figures Marique however, located the axes by finding the area of a section and its center of gravity by graphic integration The axes of inertia are the two axes, perpendicular to each other, which cross in the center of gravity and have moments of inertia greater than and less than those about any other axes through the center of gravity Koch determined the center of gravity of his sections by suspension and Grunewald who never considered the spongy bone even in sections of the femoral neck by analogy to a section with a regular geometric figure or by experimental determination of a section cut out of paper

In the moment of inertia of the various sections Koch stated that where the section is entirely composed of cancellated bone the moment of inertia is computed as for compact bone having the same size and shape The section was weighed and the ratio

of spongy bone to compact bone of the same area and thickness was computed This ratio was called the ratio of consistency of cancellous bone, and was obtained by weighing the section and then calculating the weight of the compact bone for a section of the same thickness and area The value for the ratio of consistency was then subtracted from the total weight of the section to obtain the weight of the spongy bone The effective moment of inertia in terms of compact bone, was then calculated by multiplying the ratio of consistency by the moment of inertia of compact bone Since the modulus of elasticity of compact bone was known its value could be used in the formula for computing the moment of inertia

Koch included the formula for calculating the moment of inertia in his discussion of the theory in which he states that throughout this discussion of beams it is assumed that the cross section of the beam is uniform throughout its length Such a condition however does not exist in a long bone Furthermore, in using his ratio of consistency when computing the moment of inertia for sections having both spongy and compact bone Koch tacitly assumed that the ratio of the elastic moduli was the same as the ratio of the weights In view of Rauber's (1876) findings with regard to the difference in the compressive strength of compact and spongy bone such an assumption seems unwarranted

In order to determine the moment of inertia for the section through the neck of the femur Marique also found its necessary to determine the exact proportion of compact to spongy bone The method he used was slightly different from that employed by Koch The spongy bone was carefully sawed out of the section and weighed separately from the ring of compact bone which remained Thus the proportions of spongy and compact bone in terms of the percentage of the total weight of the section were obtained It was found for the section from the neck of Femur I that the compact bone constituted 51% (0 714 gr) and the spongy bone 49% (0 686 gr) of the total weight In order to deduce the relative densities of the compact and the spongy bone the area of each was measured with a planimeter on a photograph which had been enlarged four times These

measurements showed that for the same surface, the compact bone was four times more dense than the spongy bone A coefficient of density was then calculated, from the enlarged photograph, on the basis of a diagram of the width of the section parallel to a YY axis Then, starting with the AA axis, which was perpendicular to the YY axis he took four times the length of the compact part by one time the length of the spongy part to get the coefficient of density This coefficient was then used in the formula for calculating the moment of inertia of the section

The underlying assumption for the use of the coefficient of density was that the only difference in the physical properties of spongy and compact bone was the greater density of the latter This assumption seems to be no more warranted than Koch s that the ratio of the elastic moduli of the two types of bones was the same as the ratio of their weights Furthermore Marique offered no evidence to support his contention that the weight borne by the neck of the femur was evenly distributed between the compact and spongy bone

Grunewald did not even attempt to solve the problem arising in sections having both types of bone as he stated in a footnote that herewith only the firm part of the bone (corticalis) is considered Therefore his values for stress would only apply to those sections having no spongy bone

Koch s investigation was the most thorough of the three discussed, as more sections were analyzed but Marique seemed more careful in his determinations of the principal axes of inertia In evaluating the figures for stress obtained by Koch and Marique it should be pointed out that in engineering practice bending stresses are generally determined from the formula $s = \frac{mc}{I}$ in which s is stress m the bending moment of the section c the distance from the neutral axis to the point at which the stress is determined and I the moment of inertia The formula states that the stress is directly proportional to the bending moment and the distance from the neutral axis but is inversely proportional to the moment of inertia of the cross section of the object being bent The formula was developed on the assumption that the object was composed of homogeneous isotropic material

of a uniform cross section. It was also assumed that bending occurred about one of the principal axes of inertia of the cross section and that the moduli of elasticity of the material are equal in tension and compression.

Although the cross section dimensions of a bone gradually change from one section to another, the use of the bending formula is valid. However, the fact that a bone consists of cancellous (spongy) and compact bony tissue having different physical properties invalidates the direct application of the flexure formula to the problem. In engineering practice the difficulty in calculating bending stress in beams composed of two dissimilar materials, e.g. reinforced concrete, is surmounted by equating one of the materials with an equivalent amount of the other. Such equating must be done on the basis of the relative moduli of elasticity of the materials as is revealed by the derivation of the flexure formula (see any textbook on the strength of materials). The difficulty in applying this procedure to a mathematical analysis of the bending stress in bone is that the modulus of elasticity of cancellous bone is not known.

Hulsen (1896) made eight tests of the modulus of elasticity under tension of the "pulp" from long bones and gave an average value of 2.64 kg/mm. Four of the specimens were from a human femur and four from an ox tibia. The material was obtained by extracting the organic salts with a 2% solution of hydrochloric acid. Consequently his values represent only the modulus for the organic constituents of the "pulp" rather than the complete, normal pulp.

Koch attempted to obtain the necessary factor of equivalents between cancellous and compact bone from the ratio of the weight of the two types of bone while Marique based his equivalents on the ratio of the density of spongy and compact bone. Both assumptions were false and consequently the value they obtained for the various stresses arising from bending were incorrect. The problem was especially acute in analyses of the stress in sections through the proximal and distal ends of the femur where there is a relatively large amount of spongy bone. Grunewald, as previously mentioned, made no attempt to solve the problem in dealing with such sections.

Summary

In spite of its ingenuity, several criticisms have been made of the results obtained by mathematical analysis of cross sections of bones. The first of these criticisms, as stated by Kuntscher (1934), is that the formulas used in such analyses are based on the assumption of an even distribution of force in the cross sections, a condition which rarely occurs in a bone. Although these formulas are appropriate for studying the strength of materials and forces in relatively simple bodies, they are not applicable to a complicated body such as a bone. Secondly, a bone is not an evenly formed body, whereas the theory of columns and of beams, both of which have been used in mathematical analysis of stress and strain in bones, are based on the assumption that the body being analyzed is of uniform shape throughout and composed of homogeneous material. These conditions do not apply to a bone. Thirdly, the results of mathematical studies deal with two dimensions instead of three.

Consequently, the actual values for the magnitude of the various stresses, as computed by Koch, Grunewald and Marique, produced in a femur under the assumed conditions of loading, are probably not valid. However, the results of all three investigators are in agreement that the superior aspect of the neck and the entire medial aspect of the shaft are the regions of the femur subjected to tensile and compressive strain and stress, respectively. All except Grunewald found that the highest tensile strain occurred in the superior aspect of the neck, but there was some slight disagreement as to the exact level of the medial aspect of the bone exhibiting the highest compressive strain. In each section the compressive strain and stress exceeded the tensile stress and strain for the same section.

4

Stress-Strain Studies With Models of Bones

THE FIRST attempt to study stress and strain in bones by means of models was that by Roux (1885) who used a paraffin coated rubber model of an ankylosed human knee joint. Although Roux did not know the case history of the individual from whom the specimen was obtained, he assumed that the concave aspect of the joint had been subjected to increased pressure in walking. The cracks arising in the paraffin when loads were applied to the model served as the basis for trajectorial diagrams which were then used to explain the functional significance of the orientation of the bony trabeculae across the joint. Roux believed that tensile stresses were the primary trophic stimulus for the formation of bony tissue but did not attempt to determine the magnitude of the supposed stresses and strains.

Kuntscher (1935) criticized Roux's method on the grounds that the deformations in the rubber model were so great that the plane of application of the force shifted during his tests and thus changed the direction and magnitude of the force. Neither was he convinced of the trajectorial nature of the trabeculae.

Further attempts to study stress and strain in bones by means of models were made by Hallermann (1934), Milch (1940) and Pauwels (1948, 1951). In their studies a plastic model of a bone was made and the stress patterns produced in it by various loads were studied under polarized light. The principles of this method, which is widely used for stress-strain analysis of engineering structures, were described in Chapter Two.

Hallermann used a simplified celluloid model of the leg bones and studied the stress distribution produced by even axial pressure. The pressure poles of the tibia and fibula were indicated by dark areas which disappeared under compression. After an oblique osteotomy of the tibia, continued loading resulted in

stress accumulation of the fibula diagonally opposite the osteotomy At first the fibula was under compression but a neutral line gradually appeared in the lateral edge of the model and shifted medially during the test A transverse osteotomy of the tibia was found to produce an increased compressive stress in the fibula at the level of the osteotomy The stress could be made to predominate any level by enlarging the area of the model cross section

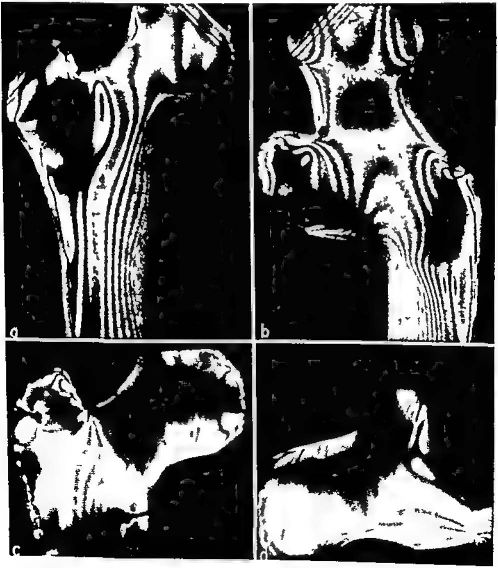

Figure 4 Photoelastic patterns in models of various bones (From Milch *J Bone & Joint Surg* 22 1940)

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Stress-Strain Studies With Models of Bones

THE FIRST attempt to study stress and strain in bones by means of models was that by Roux (1885) who used a paraffin coated rubber model of an ankylosed human knee joint. Although Roux did not know the case history of the individual from whom the specimen was obtained he assumed that the concave aspect of the joint had been subjected to increased pressure in walking. The cracks arising in the paraffin when loads were applied to the model served as the basis for trajectorial diagrams, which were then used to explain the functional significance of the orientation of the bony trabeculae across the joint. Roux believed that tensile stresses were the primary trophic stimulus for the formation of bony tissue but did not attempt to determine the magnitude of the supposed stresses and strains.

Kuntscher (1935) criticized Roux's method on the grounds that the deformations in the rubber model were so great that the plane of application of the force shifted during his tests and thus changed the direction and magnitude of the force. Neither was he convinced of the trajectorial nature of the trabeculae.

Further attempts to study stress and strain in bones by means of models were made by Hallermann (1934), Milch (1940), and Pauwels (1948, 1951). In their studies a plastic model of a bone was made and the stress patterns produced in it by various loads were studied under polarized light. The principles of this method, which is widely used for stress-strain analysis of engineering structures, were described in Chapter Two.

Hallermann used a simplified celluloid model of the leg bones and studied the stress distribution produced by even axial pressure. The pressure poles of the tibia and fibula were indicated by dark areas which disappeared under compression. After an oblique osteotomy of the tibia, continued loading resulted in

stress accumulation of the fibula diagonally opposite the osteotomy. At first the fibula was under compression but a neutral line gradually appeared in the lateral edge of the model and shifted medially during the test. A transverse osteotomy of the tibia was found to produce an increased compressive stress in the fibula at the level of the osteotomy. The stress could be made to predominate any level by enlarging the area of the model cross section.

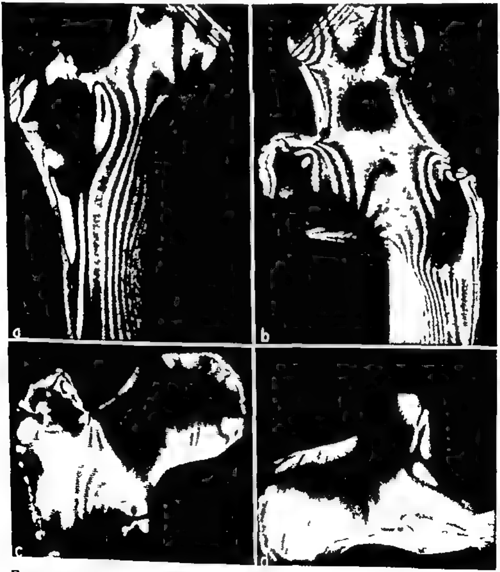

Figure 4. Photoelastic patterns in models of various bones. (From Milch, *J Bone & Joint Surg* 22, 1940.)

Hallermann's photographs are not clear, nor did he state the magnitude of the loads producing the patterns He believed the method was limited to solving problems of general bone mechanics and was not adaptable to dynamic studies because the factor of velocity could not be demonstrated

Milch published better photographs of the photoelastic patterns produced by compressive loading of models of a femur and calcaneus The stress patterns produced in models of femurs with a Lorenz bifurcation and a Schanz osteotomy were also studied The models were cut from plates of plastic $^{3}/_{8}$ of an inch thick and the stress patterns produced by loading the model, were studied under polarized light

According to Milch the stress lines produced by compressive loading of the model of a normal femur (Figure 4a) were markedly reminiscent of the trabecular structure seen on longitudinal section of the femur Tilting the model to simulate the conditions of abduction and adduction under weight bearing shifted the stress lines to one or the other longitudinal systems Longer models gave better representation of the stress lines which seemed to extend between points of compression and did not radiate into the trochanteric regions The stress lines in the normal femur model were more concentrated on the medial side toward the mechanical axis The lines usually stopped in the region of the epiphysis but this could be modified by the addition of traction forces The possible influence of the tension exerted by the trochanteric muscles was investigated by studying the stress pattern (Figure 4b) produced by pulling on wires simulating muscles attached to the trochanters of a model femur

In a model of a femur with a Lorenz bifurcation the stress pattern produced by loading the upper end of the proximal segment was concentrated along the medial aspect of the shaft However if the load were applied to the head of the bone the stress lines shifted to the lateral aspect of the shaft in the direction of the misplaced mechanical axis Similar results were obtained from loading a model of the femur having a Schanz osteotomy (Figure 4c) With two point loading the majority of the stress lines were longitudinally oriented along the lateral border of the shaft Evidence of stress appeared in the base of the

greater trochanter but few if any lines were seen in the lesser trochanter Stress patterns were also obtained by compressive loading of a model of a normal calcaneus (Figure 4d) The load was applied at three points to simulate normal weight bearing

Milch believed that the stress patterns emphasized the importance of the mutual relations of the anatomical and mechanical axes as related to the deposition of bony tissue Thus, shifting of the mechanical axis to either side of the anatomical axis was accompanied by a similar change in the position of the stress lines In an actual femur according to Milch, the aspect of the bone corresponding to the region of stress concentration in the model would be thickened by increased deposition of bony tissue Thickening of either the lateral or the medial aspect of the femoral shaft would tend to convert the bone into a straight rod This is prevented by lengthening of the femoral neck Bending seemed to have little influence on the production of the stress pattern

Milch assumed that the stress lines in his models did or could be made to represent actual conditions of loading in real bones He also believed that the stress patterns were experimental confirmations of both Roux s and Wolff s laws of bone growth and transformation Both of these authors were adherents of the trajectorial theory of bone architecture, according to which the trabeculae of spongy bone are laid down along the lines of maximum internal stress trajectories in the bone Consequently the lines of trabeculae should cross one another at right angles and should be perpendicular or tangent at their points of junction with the margin of the bone or articular cartilage However, the stress lines in the photoelastic patterns Milch obtained do not conform with these requirements

Roux and Wolff also believed that tensile stresses were the primary stimulus for bone growth but Milch made no clear differentiation regarding the type of stress represented in his photo elastic patterns The majority of the stress lines in the various models illustrated were more or less parallel and did not in the author s opinion seem reminiscent of the trabecular orientation of actual bones

In contrast to Hallermann and Milch the photoelastic tech-

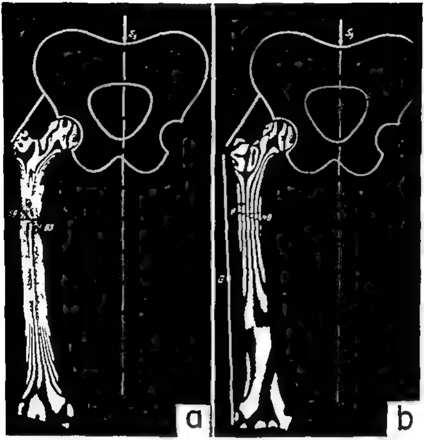

Figure 5 Photoelastic patterns in models of a femur. (From Pauwels Acta Anat 12 1951)

nique was used by Pauwels (*loc cit*) to demonstrate the way in which muscles and ligaments acting as traction braces reduce the magnitude of the stress in the femur and humerus respectively The numerals along the sides of the illustrations of the stress patterns Pauwels obtained in his studies are the fringe numbers which represent the order of appearance of the dark bands in the models Judging by the values he gave for the stress the thickness of the models he used was such that the stress was equal to ten times the fringe number

In the model of a femur without a traction brace (Figure 5a) the body weight alone produced a tensile stress of 69 kg/cm² on the lateral aspect of the shaft and a compressive stress of 83 kg/cm² on the medial aspect of the shaft When a traction brace

representing the iliotibial tract, was attached to the greater trochanter of the model femur the highest tensile stress was reduced to 8 kg/cm² and the compressive stress to 48 kg/cm² (Figure 5b) The reason for this reduction in the magnitude of the stress is that the traction brace tends to bend the femur laterally, thus reducing the medial bending produced by the body weight

The magnitude of the tensile and compressive stress produced in the anterior and posterior aspect respectively, of the humeral shaft by the weight of the forearm when flexed to a 90 angle showed considerable variation in the proximal and distal half of the model (Figure 6a) In the proximal half of the model the highest marginal tensile and compressive stress was 85 kg/cm² and 81 kg/cm² respectively but in the distal half of the model

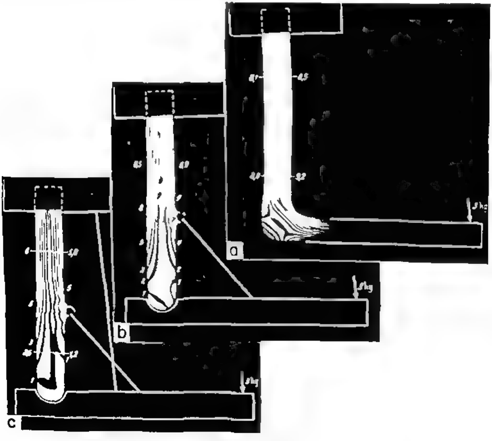

Figure 6 Photoelastic patterns in models of a humerus (From Pauwels *Acta Anat* 12 1951)

the corresponding stresses had increased to 92 kg/cm^2 and 88 kg/cm^2.

When a single traction brace, representing a uniarticular muscle like the brachialis, was added the magnitude of the stresses was reduced in a proximo distal direction (Figure 6b). Thus the tensile stress was reduced from 89 kg/cm^2 in the proximal half of the model to 10 kg/cm^2 near the articular surface at the distal end of the shaft. The compressive stress on the opposite or posterior aspect of the shaft was decreased from a maximum of 85 kg/cm^2 in the proximal half to a minimum of 10 kg/cm^2 near the distal articular surface. The stress in the part of the humerus spanned by the muscle is reduced because that portion of the bone is bent in a direction opposite to that produced by the weight of the forearm. Furthermore the lever arm through which the muscle acts upon the bone, represented by the vertical distance between the action line of the muscle and the anatomical axis of the bone, increases toward the joint. This enhances the mechanical efficiency of the muscle. The effect of these various factors is to decrease the stresses toward the joint. Thus the magnitude of the stress is least at the center of the joint where the bending action produced in the humerus by the weight of the forearm is almost eliminated.

With the addition of a second traction brace representing a biarticular muscle such as the biceps brachii the stresses were reduced evenly throughout the entire length of the humerus (Figure 6c). Thus the tensile stress in the anterior aspect of the humerus decreased from 89 kg/cm^2 to 58 kg/cm^2 in the upper part and from about 20 kg/cm^2 to 12 kg/cm^2 in the lower part of the shaft. The compressive stress in the posterior border of the humerus was reduced from 85 kg/cm^2 to 69 kg/cm^2 in the proximal half and from about 30 kg/cm^2 to 25 kg/cm^2 in the distal half of the region. The explanation for this relatively even reduction in the stress of the different parts of the model lies in the fact that the traction brace is only slightly inclined to the anatomical axis of the model humerus. Consequently there would be relatively little increase in the length of the lever arm through which the brace (or muscle) acts upon the humerus.

Summary

The chief criticism against analysis of stress and strain in bone by means of the photoelastic technique is that the method, as used in engineering, is for stress strain analysis of solid objects of homogeneous composition. In all the works cited above the bone models studied were made from a solid resinous material of quite uniform composition. Furthermore, they were usually cut from thin plates of the material so that the resulting stress pattern was in just two dimensions. All of these models were quite different from a hollow bone of heterogeneous composition and complicated trabecular organization. The latter feature, as Kuntscher (1934) pointed out, is impossible to duplicate in any model. Thus, the various photoelastic patterns obtained by Hallermann, Milch, and Pauwels simply represent the stresses produced by loading solid models composed of homogeneous material resembling a bone in outline.

5

Stress and Strain in Skull Deformation and Fracture

STRESS AND STRAIN in the human skull has been experimentally studied under both dynamic and static conditions in attempts to determine the strength of skull bones and the mechanism of skull fracture. In dynamic loading the force or load is suddenly applied to the skull, while in static loading it is slowly applied. Most skull or head injuries occur under dynamic conditions as the result of impacts and are energy problems. Therefore, the results obtained from dynamic loading studies on skull deformation and fracture will be considered first.

Dynamic Loading Studies

The behavior of the intact human skull under dynamic loading has been studied extensively by Gurdjian, Lissner, and their associates with the aid of the stresscoat method and electric strain gages. The principles of these methods are discussed in Chapter Two.

The first problem investigated was how closely the stresscoat deformation patterns in dry bones resemble those which would be produced in living bones under similar experimental conditions. The problem was solved (Gurdjian and Lissner 1945) by comparing the strain pattern produced in the stresscoated skull of six dogs and two macaque monkeys under three different experimental conditions. In the first series of tests the scalp of an anesthetized animal was reflected, the underlying skull cleaned, stresscoated, and then hit with a hammer. In the second series of tests the animal was sacrificed and the experiment repeated with the skull contents intact. In the third series of tests the experiment was repeated on the thoroughly cleaned and dried skull of the same animal.

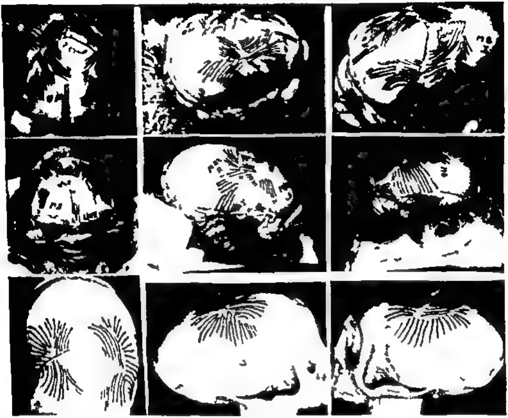

Figure 7 Stresscoat patterns on a monkey skull (From Gurdjian and Lissner *Surg Gynec & Obst* 81 1945) Upper row—Pattern on skull of anesthetized monkey Middle row—Pattern on skull of dead animal with skull contents intact; Bottom row—Pattern on dry skull of the same animal

These studies clearly demonstrated that the strain pattern (Figure 7) was essentially similar regardless of the experimental conditions Therefore conclusions based on stresscoat studies of a dry *bone are valid for predicting the tensile strain pattern* which would be produced in the corresponding living bone under similar experimental conditions The chief differences noted in the tests were that the strain pattern was somewhat more extensive in the skull of the living animal and that in the dry skull the propagation of the strain tended to be interrupted by the suture lines The resemblance between the deformation pattern of the living and of the dry skull indicates that the direction of the tensile strain was similar in both but does not imply that the magnitude of the strain was the same

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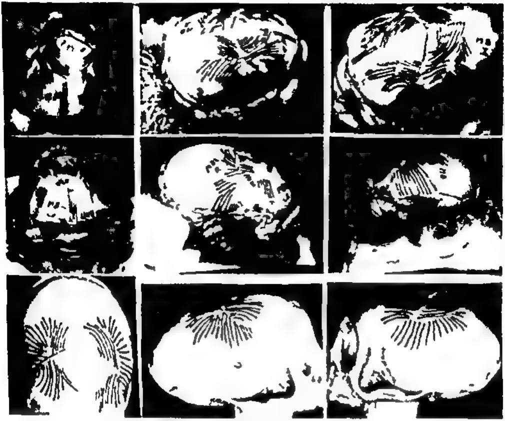

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The strain patterns revealed that striking the skull with a hammer caused an inbending of the bone, which gave rise to tensile strain in a circular and a radial direction around the site of impact Radial strain, indicated by concentrically oriented stresscoat cracks was closer to the site of impact Circular strain shown by radially oriented cracks was more distant Similar tensile strain occurs in a rubber ball when pushed in with the thumb

After these experiments tensile strain patterns were produced in human skulls by hammer blows of varying intensities Five dry skulls and three intact cadaver heads with the scalp reflected so the bones could be stresscoated were used In contrast to the rather generalized and extensive patterns seen in the dog and monkey skulls discrete patterns at the site of impact were found Usually only radially oriented cracks indicating tensile strain in a circular direction around the site of impact were present Superimposition of several strain patterns indicated that after a blow the skull vibrated before coming to equilibrium and rest

The most important result of the study was the demonstration that linear fractures of the skull arose from failure of the bone as the result of the tensile stress created in it by the blow Frequently the most extensive deformation occurred some distance from the point of impact of the blow The shape and thickness of the skull as well as the area to which the blow was applied influenced the *development of the deformation pattern* These factors account for the differences between the tensile strain patterns in the human skull and those of the dog and the monkey

Gurdjian and Lissner (1946) also made quantitative determinations of deformations experimentally produced in human cadaver skulls The skulls were kept in a saline solution until tested at which time they were coated with stresscoat lacquer having a sensitivity of 0 00100 ± 0 0005 inches/inch Deformations produced by application of 14 3 inch pounds of energy to different regions of the skull were studied in six specimens weighing from 1 07 to 2 23 pounds The energy was applied by dropping the skulls on a 140 lb steel block in such a way that the blow was perpendicular to a plane tangent to the point of impact on the skull Each skull was caught by hand on the rebound so

that it struck the steel block just once The weight of the skull multiplied by the distance through which it was dropped gave the energy (inch pounds) applied to the skull The velocities of the skull at the time of impact varied from 5 4 to 9 9 feet per second

In the first six experiments 14 3 inch pounds of energy were applied to all the skulls but later varying amounts of energy were used The strain patterns produced by a constant amount of energy varied in different places of a single skull as well as between skulls Three skulls were tested with energies varying from eight to 24 inch pounds, applied to the midfrontal, the mid occipital and the lateral posterior parietal regions In these skulls it was found that eight inch pounds of energy in the oc cipital region ten inch pounds of energy in the lateral posterior parietal region, and 14 to 28 inch pounds of energy in the mid frontal region produced a threshold or minimal tensile strain pattern The path of the strain depended upon the shape contour, and thickness of the bone in the area of the impact The deformation patterns were more common in the weaker than in the buttressed regions of the skull relatively few patterns being found in the region of the zygomaticofrontal and petrosoparietal buttresses There were also a few cracks in the immediate plane of the frontal and of the occipital bones The results of these experiments were corroborated by clinical data The ease with which deformations were produced in the region of the foramen magnum and in the parietotemporal area may, according to the authors account for the many fractures of these regions which occur without loss of consciousness

The tensile strain pattern produced on the outer and the inner aspects of the skull by application of 16 inch pounds of energy to various regions was likewise investigated (Gurdjian, Lissner and Webster, 1947) A stellate pattern was found on the inner surface of the skull opposite the point of impact while on the outer aspect of the skull the stresscoat cracks radiated from around the general area of the impact (Figure 8) The stellate pattern consisted of radially arranged cracks indicating tensile strain in a circular direction as the result of the inbending of the bone The cracks on the outer aspect of the skull, lying some

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sion fractures or perforations, the deformation patterns suggest that shearing may have occurred between the outer and the inner table of the skull

The linear fractures produced in the tests arose from failure of the bone because of the tensile strain in the outer surface of the skull The fracture was initiated some distance from the site of impact and then propagated both toward and away from the

Figure 9 Undulation of the skull at impact (From Gurdjian Webster and Lissner *Am J Surg* 78 1949)

Figure 8 Stresscoat patterns on human skull (From Gurdjian Lissner and Webster *Surg Gynec & Obst* 85 1947)

distance from the site of impact, arose from outbending of the bone and signified a tensile strain in a more or less circular direction around the area of impact The stellate pattern suggested that the tensile strain was concentrated at the site of impact and inbending of the skull

Although the stresscoat pattern only shows areas of tensile strain it is known that when bending occurs tensile strain is created on the convex surface and compressive strain on the concave surface of the bent bone Therefore, the presence of a deformation pattern on just one surface of a skull indicates that bending had occurred in the direction of the aspect of the skull having the pattern Inbending of the skull is accompanied by outbending in certain selected areas showing that an undulation of the skull had occurred (Figure 9) The inbending is in the region of the impact while the outbending starts at the border of the inbent area After a deceleration impact the inbent area is limited by the radius or curvature of the skull in the region of the blow Because of differences in the shape and thickness of the skull a single impact may produce outbending in different planes of the various regions of the skull The inbent area is also modified by the presence of buttresses and foramina When the blows are not sufficiently severe or localized to produce depres

mation without fracture than can the outer table The study clearly proved that if the site of impact is known the direction of the resulting fracture line can be surmised and vice versa

Skull deformations and changes in intracranial pressure occurring in dogs at the time of head injury have been measured by Gurdjian and Lissner (1944), with wire resistance strain gages (Figure 10) and cathode ray oscilloscopes Small plastic pressure plugs traversed by platinum wire electrodes in contact with the brain and cerebrospinal fluid, were placed in the temporal regions of the skull The resistance changes in the strain gage (Sr4 Type C-5) were measured by a potentiometer circuit and recorded on an oscilloscope

Application of a blow close to a strain gage produced compressive strain (Figure 11a) because of the inbending of the bone Tensile strain arising from an outbending of the skull or a decrease in its radius of curvature occurred when the blow was applied to the bone on the side opposite to that bearing the gage (Figure 11b) Approximately 1/1000 of a second was required for maximum deformation of the skull After its initial deformation the skull returned to its original shape although two to four additional oscillations of deformation occurred Equilibrium was attained in about 1/1000 of a second after the beginning of the blow Increasing the intensity of the blow increased the amplitude of the deformation but not the time period of the initial deformation which remained 1/1000 of a second (Figure 11c) How

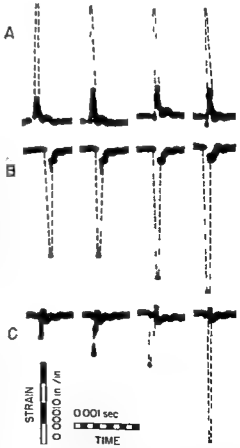

Figure 11 Oscillograph records of deformation in dog skull (Redrawn from Gurdjian and Lissner *J Neurosurg I* 1944)

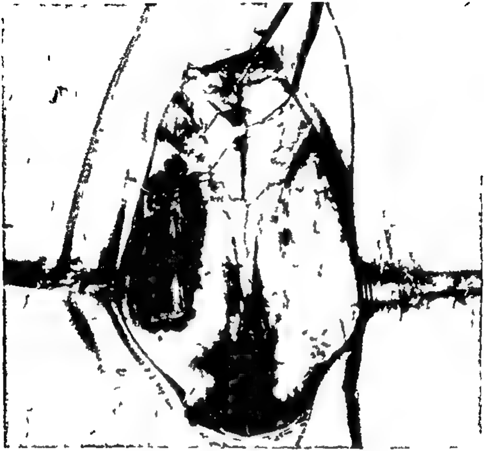

Figure 10 Electric strain gage on dog skull (From Gurdjian and Lissner *J Neurosurg* 1 1944)

impact site The fracture line was propagated toward the site of impact because at the instant of impact the area of inbending (the outer surface of the skull at the site of impact) is actually under compression The rebounding of the bone creates tensile strain with its greatest magnitude at the center of the impact area From this center point outward the tensile strain decreases rapidly in all directions On the basis of the stellate pattern on the inner surface of the skull the fracture should extend in both directions from the center of the area of impact as far as the border of the area of inbending but linear fractures did not always occur The latter feature may be related to the two layer construction of most of the vertex of the skull the inner table because of its thinness, being able to undergo a greater defor-

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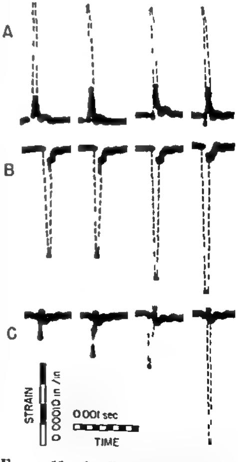

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ever, the time period was dependent upon the size and shape of the skull.

The intracranial pressure changes were similar to the skull deformations. A hammer blow on the same side as a pressure plug caused an initial pressure increase in the region of the plug followed by two to four pressure oscillations before equilibrium was re-established. Simultaneously there was an initial decrease in intracranial pressure near the plug on the side of the skull opposite the blow. Striking the interparietal region of the skull produced simultaneous increase in pressure on both sides. An average of about $\frac{1}{400}$ of a second was required to reach the initial pressure peak. This is different from that required to reach maximum skull deformation because the skull generally does not attain maximum acceleration until after maximum deformation has occurred. Therefore, the deformation and pressure waves, which were not in phase, dampened each other and restored deformation and pressure equilibrium in a short time.

In a later study with the strain gage cathode ray oscilloscope method, Gurdjian and Webster (1947) found that striking the skull of an anesthetized dog produced deformations similar to those obtained 24 hours after death of the animal and with the skull contents intact. They, therefore, concluded that cadaver heads with the skull contents intact were suitable for investigating deformations arising from blows on the living head. They found that the initial deformation from a blow on a human cadaver skull varied from $\frac{1}{5000}$ to $\frac{1}{1000}$ of a second. The total elapsed time for the appearance of the deformation pattern was from $\frac{1}{200}$ to $\frac{1}{250}$ of a second. During the deformation three to six cycles, with frequencies from about 900 to 1200 cycles per second, occurred in different parts of the skull.

The amount of energy and the time of absorption required to fracture the skull have been investigated by Gurdjian, Webster and Lissner (1949) in 55 intact human cadaver heads. Impact tests were made by dropping the head on a steel block so that energy was applied to the mid-frontal, anterior interparietal, mid-occipital, and the right or the left posterior parietal region.

The energy required to produce a single linear fracture varied from 400 to 900 inch pounds. The average energy necessary for

fracture was 571 inch pounds in the mid frontal regions, 517 inch pounds in the mid occipital region, 710 inch pounds in the anterior interparietal region, and 615 inch pounds in the region above each ear However, because of the wide range of variation in each region e g 425 to 803 inch pounds in the mid frontal region the average differences are of little importance After the initiation of a single linear fracture very little more energy was required to produce multiple fractures and complete destruction of the skull The average energies for complete destruction of the skull were close to those producing single linear fractures, and it is significant that fracture sometimes occurred with approximately 400 inch pounds of energy

The great variation between skulls in the amount of energy required for fracture was the result of individual variations in the shape and thickness of the skull and scalp One of the most important results of the study was the demonstration that a relatively thin layer of soft tissue, the scalp, was an excellent energy absorbing material Thus, 400 or more inch pounds of energy were necessary to produce a linear fracture in an intact cadaver head while a similar type of fracture occurred in a dry skull with as little as 40 inch pounds of energy

The authors believe that the difference between the amount of energy necessary to fracture a cadaver head and that of a living individual is of the same order of magnitude, because of variations in such anatomical factors as scalp thickness This belief is based on the amount of kinetic energy producing a skull fracture when a batter is accidentally hit on the head by a fast ball The energy was computed by using 100 feet per second as the speed of the ball and five ounces as its weight giving a value of 580 inch pounds of energy

One of the important factors in skull fracture is the time of absorption of the energy producing the fracture as a large amount of energy can be withstood without fracture if it is absorbed slowly Gurdjian Webster and Lissner (1950a) determined the time of absorption of the energy in cadaver heads by mounting Sr-4 electric strain gages along the paths of the expected fracture lines in the skull The head was dropped upon a steel block so that it struck a switch connected to one beam of a

two gun cathode ray tube, which indicated the initial contact of the skull with the steel block The other beam was controlled by a similar strain gage connected to a simple potentiometer circuit Fortunately, in one skull the fracture passed through the strain gage thus opening the circuit The elapsed time from the instant the skull struck the steel block until it began to deform was 0 0006 of a second but the actual fracture did not occur until the bone itself had been deformed through an additional period of 0 0006 of a second Thus the time from the instant the skull struck the steel block until the fracture occurred was 0 0012 of a second

Gurdjian Webster, and Lissner classified depression skull fractures into six types on the basis of velocity, kinetic energy, and shape of the injuring object A very high velocity object such as a rifle bullet perforates the skull producing shattering of the bone from the radial acceleration and great increase in intracranial pressure

A second type of fracture is caused by a pistol bullet or some other fairly high velocity object The skull will also be perforated and bone fragments may be forced into the brain If the energy is not too high only the outer table of the skull is depressed or perforated since most of the energy is dissipated at the time of impact

Blunt objects with a much lower velocity than bullets e g a baseball produce the third type of depression fracture in which most of the energy is absorbed in producing the depression The latter is an oval inbent area with radial fractures and separation of the two tables of the skull A curvilinear fracture caused by tensile stress on the external surface of the skull is usually present at the border of the depressed area The fragmentation of the depressed area arising from tensile stresses due to inbending of the bone roughly corresponds to the stresscoat pattern found on the inner surface of the skull around the site of impact If the energy of the striking object is almost completely dissipated at the time of impact either skull table depending on which is the thicker may be depressed

A localized blow from a slowly moving object produces the fourth type of depression fracture in which simultaneous defor

mation in other regions of the skull may occur In addition, one or two linear fractures, arising from tensile stresses produced by distant outbending of the bone, extend toward the site of impact

In the fifth type the depression caused by a slowly moving pointed object, may correspond to the shape of the object Linear fractures arising from distant deformations may occur The sixth type exhibits extensive comminution with radial and circular fracture lines The former extend from the center of the impact site while the latter surround the area at varying distances This type of fracture is produced by a slowly moving blunt object, with high kinetic energy, and frequently accompanies a deceleration impact

The same investigators (Gurdjian Webster, and Lissner, 1950 a and b) continued their researches on the mechanism of skull

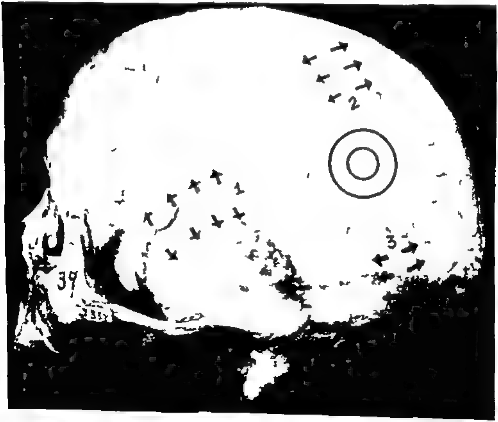

Figure 12 Areas of different stress level following posterior parietal impact (From Gurdjian Webster and Lissner *J Neurosurg* 7 1950)

two gun cathode ray tube which indicated the initial contact of the skull with the steel block. The other beam was controlled by a similar strain gage connected to a simple potentiometer circuit. Fortunately, in one skull the fracture passed through the strain gage, thus opening the circuit. The elapsed time from the instant the skull struck the steel block until it began to deform was 0.0006 of a second, but the actual fracture did not occur until the bone itself had been deformed through an additional period of 0.0006 of a second. Thus, the time from the instant the skull struck the steel block until the fracture occurred was 0.0012 of a second.

Gurdjian, Webster, and Lissner classified depression skull fractures into six types on the basis of velocity, kinetic energy, and shape of the injuring object. A very high velocity object, such as a rifle bullet, perforates the skull, producing shattering of the bone from the radial acceleration and great increase in intracranial pressure.

A second type of fracture is caused by a pistol bullet or some other fairly high velocity object. The skull will also be perforated and bone fragments may be forced into the brain. If the energy is not too high, only the outer table of the skull is depressed or perforated, since most of the energy is dissipated at the time of impact.

Blunt objects with a much lower velocity than bullets, e.g., a baseball, produce the third type of depression fracture in which most of the energy is absorbed in producing the depression. The latter is an oval inbent area with radial fractures and separation of the two tables of the skull. A curvilinear fracture caused by *tensile stress on the external surface of the skull is usually present* at the border of the depressed area. The fragmentation of the depressed area, arising from tensile stresses due to inbending of the bone, roughly corresponds to the stresscoat pattern found on the inner surface of the skull around the site of impact. If the energy of the striking object is almost completely dissipated at the time of impact, either skull table, depending on which is the thicker, may be depressed.

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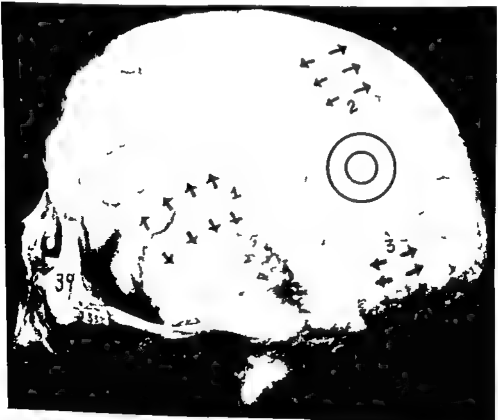

Figure 12 Areas of different stress level following posterior parietal impact (From Gurdjian Webster and Lissner *J Neurosurg* 7 1950)

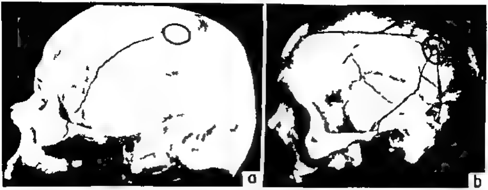

Figure 13 Single (a) and multiple (b) linear fractures from a posterior parietal blow (From Gurdjian Webster and Lissner *J Neurosurg* 7 1950)

fracture by analyzing the stresscoat pattern produced by blows in each of 12 different regions By noting the time of appearance of stresscoat cracks produced by increasing amounts of energy, they were able to determine areas of primary, secondary and tertiary stress levels (Figure 12) Thus the first set of cracks appearing after an impact denoted primary stress level the second set of cracks arising from additional energy indicated the secondary stress level while still more energy produced a third set of cracks showing the tertiary stress level

The significance of these stress levels was then related to fractures experimentally produced by applying varying amounts of energy to each of 12 areas of the skull Embalmed cadaver heads with scalp and skull contents intact were used In each specimen the thickness of the scalp was measured and after testing the skull was cleaned weighed and the fracture photographed Forty six cadaver heads were tested with deceleration impacts applied to the mid frontal the anterior parietal the mid-occipital and the posterior parietal regions

The resulting fractures agreed with the predicted results from stresscoat patterns obtained by loading corresponding areas of the dry skull Thus a single linear fracture (Figure 13a) appeared in the area showing primary tensile stress level If two fractures were present the second one appeared in the area of secondary stress level and if there were three linear fractures the third arose in the area of tertiary stress level The area of

primary stress level is the weakest region of the skull, but the actual location of this area, following an impact in a similar area of different specimens, may vary.

In the experimentally produced fractures there were sometimes as many as six fracture lines extending radially from the site of impact (Figure 13b). The fractures arose from failure of the bone because of the tensile stresses on the internal surface of the skull. Failures on the external aspect of the skull occurred at the junction of the inbent and the not yet inbended area. In general, the more rapid the blow the more localized the inbent area and the less distant the outbent area. Propagation of the fracture lines was not hindered by skull buttresses parallel with the direction of the tensile stress. The point of application of the energy determined the general region and direction of the resulting fracture. Evidence from the stresscoat patterns and the experimental fractures enabled the investigators to predict the location of primary and secondary fracture lines when the point of application of the blow was known. However, because of the brittleness of the skull, it was very difficult to foretell the order of appearance of additional fracture lines in the stellate pattern.

The results of the stresscoat tests were substantiated by a study of clinical cases of head injury. Thus, in cases where the

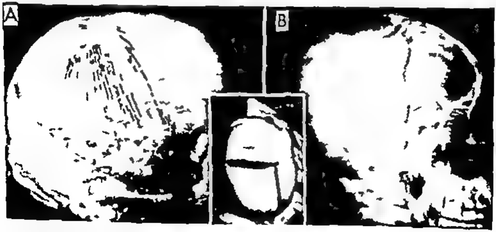

Figure 14. A stresscoat pattern (A) produced by an anterior interparietal blow compared with a similarly produced clinical fracture (B). (From Gurdjian, Webster and Lissner, *Radiology* 54, 1950b.)

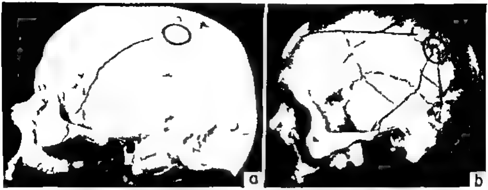

Figure 13. Single (a) and multiple (b) linear fractures from a posterior parietal blow. (From Gurdjian, Webster and Lissner, *J. Neurosurg.* 7, 1950.)

fracture by analyzing the stresscoat pattern produced by blows in each of 12 different regions. By noting the time of appearance of stresscoat cracks produced by increasing amounts of energy, they were able to determine areas of primary, secondary, and tertiary stress levels (Figure 12). Thus, the first set of cracks appearing after an impact denoted primary stress level, the second set of cracks arising from additional energy indicated the secondary stress level, while still more energy produced a third set of cracks showing the tertiary stress level.

The significance of these stress levels was then related to fractures experimentally produced by applying varying amounts of energy to each of 12 areas of the skull. Embalmed cadaver heads with scalp and skull contents intact were used. In each specimen the thickness of the scalp was measured, and after testing the skull was cleaned, weighed, and the fracture photographed. Forty-six cadaver heads were tested with deceleration impacts applied to the mid frontal, the anterior parietal, the mid occipital, and the posterior parietal regions.

The resulting fractures agreed with the predicted results from stresscoat patterns obtained by loading corresponding areas of the dry skull. Thus, a single linear fracture (Figure 13a) appeared in the area showing primary tensile stress level. If two fractures were present, the second one appeared in the area of secondary stress level; and if there were three linear fractures, the third arose in the area of tertiary stress level. The area of

as Aran believed in 1844 Neither do fractures always arise from flattening of a curved surface as stated by Felizet in 1873, or follow a skull buttress The concept of Rawling (1904) that basal fractures originate from an actual splitting of the skull was also not verified

Static Loading Studies

The first extensive investigation on the breaking strength of intact bones was made by Messerer (1880) Approximately 500 fresh human bones were statically loaded to failure in a Werder sche testing machine The orientation of the bones in the testing machine was varied so as to change the direction of the force with reference to the axes of the bone The deformations or changes in dimensions of the bone occurring during a test were measured in mm

The breaking strength of intact male and female skulls was studied by applying the pressure in a side to side or transverse direction and in a front to back or longitudinal direction The skulls of seven males varying from 18 to 69 years of age and of six females ranging from 22 to 82 years of age, were tested in transverse loading The average age was 42 7 years for the males and 48 [illegible] years for the females The average changes in the transverse, longitudinal and perpendicular dimensions of the skull with transverse pressure are tabulated below

Total Change to Fracture

Male		Female	
4 25 mm	Transverse decrease	5 66 mm	(5 skulls)
4 285 mm	Longitudinal increase	0 464 mm	(5 skulls)
7 48 mm	Perpendicular increase	0 688 mm	(5 skulls)
	Changes in the Presumed Elastic Limit		
1 657 mm	Transverse decrease	2 36 mm	(5 skulls)
0 168 mm	Longitudinal increase	0 128 mm	(5 skulls)
0 168 mm	Perpendicular increase	0 215 mm	(4 skulls)
	Elastic Changes for 100 kg Load		
6 7 mm	Transverse decrease	8 60 mm	(5 skulls)
0 06 mm	Longitudinal increase	0 05 mm	(5 skulls)
0 068 mm	Perpendicular increase	0 09 mm	(4 skulls)

Seven skulls of males and five of females were tested with the pressure applied in the longitudinal direction The average

site of the blow was known, the resultant fracture occurred as predicted from the stresscoat experiments (Figure 14).

The studies on the prediction of fracture site in head injury were summarized (Gurdjian, Webster and Lissner, 1953) on the basis of the results of tests on 100 randomly selected adult skulls. Deceleration blows were given to the mid frontal, the anterior interparietal, the posterior interparietal, the mid occipital, the left frontal, the left anterior parietal, the left posterior parietal and the left parieto-occipital areas. The frontal, anterior parietal, posterior parietal and parieto-occipital areas were bi-lateral in position. Because of symmetry the left side and mid line areas were used, with special care taken that every part of each area was tested many times. The fractures produced by tests of each area were then analyzed according to the percentage of fractures with respect to site and direction.

In general it was found that frontal blows resulted in more vertically oriented fractures or ones slanting slightly forward, while blows in the anterior and the posterior parietal and in the parieto-occipital area gave rise to more horizontal fractures. Many fractures on one side or the other were initiated in the temporo-parietal areas. A blow on one or the other side of the mid line gave rise to ipsilateral fractures. When a blow was applied near the border of an area the fracture position in the adjacent area aided in predicting the location of the fracture arising from the blow. Thus a horizontal fracture from supero-inferior stresses in the temporal and lower parietal regions arose from blows in the anterior and inferior regions of the left parieto-occipital area.

Lateral blows resulted in transverse basal fractures unassociated with fractures of the vertex of the skull in just 2% of the tests. However, they occurred in 5% of the cases when the blow was applied to the occipital region.

The stresscoat studies of the skull clearly demonstrate the importance of tensile stress and strain in skull fractures but disagree with some of the ideas as cited by Gurdjian, Webster and Lissner (1950a) of previous investigators. Fractures of the skull vault, from blows on the vertex, do not reach the skull base by the shortest possible route or start at the site of impact

and that the skull never completely returned to its original shape after removal of the load He believed that pressure in one direc tion produced tensile forces in parts of the skull perpendicular to the direction of the pressure Consequently, there was a de crease in the radius of curvature of the part of the skull involved and eventual cracking in the area of outbending

Some of the earlier concepts on skull fracture were substan tiated by LeCount and Apfelbach (1920) and are included by Rowbotham (1942) in his discussion on the mechanism of skull fracture However, the results obtained by static loading are so different from the dynamic loading which generally occurs in head injury and fracture that they cannot be used to interpret the behavior of the skull under impact

The investigations of Gurdjian Lissner and Webster showed that all blows on the head produced both local and generalized deformations In linear fractures which arise from failure under tensile strain the local deformation from the impact is elastic and the bone rebounds after the blow In depression fractures there is a local failure of the bone at or around the site of im pact All their work on the biomechanics of skull fracture has been reviewed by Gurdjian, Webster, and Lissner in an article in Medical Physics, Vol II 1950

Split line Studies

Recently the facial part of the human and chimpanzee skull was investigated by Tappen (1953), who used the split line tech nique of Benninghoff in an attempt at functional analysis of the orientation of the Haversian systems The split-line patterns (Figure 15) he obtained were interpreted in terms of stresses and strains, presumably arising from muscle activity in chewing However stress and strain were not differentiated from one an- other, and no attempt was made to determine their nature or magnitude

In discussing the influence of muscle action upon the skeleton inappropriate analogies were used For example the zygoma was compared to a uniformly loaded horizontal beam supported at both ends and the longitudinal arrangement of the Haversian systems of the zygoma as revealed by the split line pattern

age of the males was 41.4 years (19-61) and the females 41.8 years (20-74). The average changes in the transverse, longitudinal and perpendicular dimensions of the skull are summarized as follows:

Total Change to Fracture

Male		Female
2.77 mm	Longitudinal decrease	2.76 mm
0.39 mm	Transverse increase	0.33 mm
0.18 mm	Perpendicular increase	0.19 mm
(5 skulls)		

Changes to the Presumed Elastic Limit

0.97 mm	Longitudinal decrease	1.32 mm
0.13 mm	Transverse increase	0.16 mm
0.84 mm	*Perpendicular increase*	0.13 mm
(5 skulls)		(4 skulls)

Elastic Changes for 100 kg. Load

0.31 mm	Longitudinal decrease	0.44 mm
0.41 mm	Transverse increase	0.54 mm
0.20 mm	Perpendicular increase	0.0375 mm
(5 skulls)		

From the above results it is evident that the changes to fracture with side to side pressure are not the same in the two sexes. The female skulls had a greater decrease in the transverse diameter accompanied by less increase in the longitudinal and perpendicular diameters than did the male skulls. With pressure in a longitudinal direction the total changes to fracture were almost the same in both sexes, the longitudinal decrease and transverse increase being very slightly greater in the male skulls while the female skulls showed a little greater increase in the perpendicular diameter. The average load required to fracture female skulls by side to side pressure was 562.5 kg., while the male skulls failed with an average load of 489.28 kg. With pressure in the longitudinal direction (front to back) the situation was reversed, the male skulls fracturing with an average of 685.7 kg. and the female skulls with an average of 610 kg.

Bruns, cited by Gurdjian, Webster and Lissner (1950a), in 1854 also studied skull deformation and fracture by static loading. He reported that compression decreased the diameter of the skull in one direction while increasing it in an opposite one.

gives considerable resistance to the shearing forces exerted by the muscle. The muscle involved is the masseter, whose fibers, especially those of the deeper part of the muscle, have a direction almost perpendicular to the long axis of the zygoma. Consequently, because muscle fibers always pull in a straight line, the action line of the masseter is also nearly perpendicular to the long axis of the zygoma. Therefore the contraction of the masseter muscle would tend to bend the zygoma inferiorly which would produce tensile and compressive, as well as shearing stresses within it. The tensile and compressive stresses would be greatest in the middle of the zygoma, and the shearing stresses greatest at the ends of the zygoma, where it is continuous with the rest of the skull. There is nothing in the orientation of the split lines to indicate these differences. The arguments advanced by Tappen in support of his interpretations of the split line patterns were largely theoretical and, as he admitted, there was no experimental evidence to supplement them. In the author's opinion the functional significance of the split line patterns in bones, especially with reference to stress and strain, still awaits experimental confirmation.

Summary

Stresscoat studies of the human skull under dynamic loading show that all blows on the head produce both local and generalized deformations. Tensile strain and stress is more dangerous than compressive stress and strain because linear fractures of the skull arise from failure of the bone as the result of tensile stresses within it. In linear fractures the local deformation from the impact is elastic and rebounding of the bone occurs. In depression fractures local failure of the bone occurs at and around the site of impact.

Most head injuries arise from impacts and are dynamic problems involving energy transfer and absorption. Therefore the results of static loading studies of the skull are of little or no help in understanding the behavior of the skull in head injuries.

The results of split line studies as a means of explaining skull architecture in terms of stresses and strains are questionable.

Figure 15 Split line pattern in human (a) and chimpanzee (b) skull (From the original Figures 1 & 2 Tappen *Am J Phys Anthropol* *11* 506 & 508 1953)

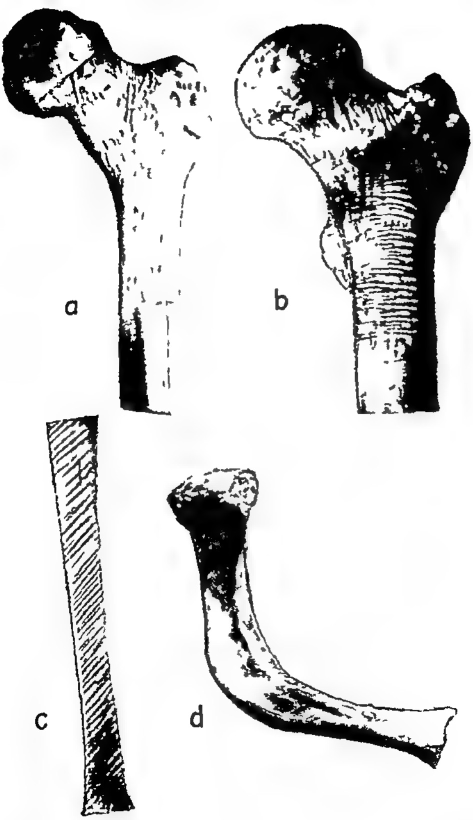

Figure 16 Tensile strain patterns in colophonium coated bones (From Kuntscher a b and c *Zentralbl f Chir* 61 1934 . . . f *Klin Ch* 182 1935)

6

Stress-Strain Distribution in Long Bones

Results With the Colophonium Method

THE GERMAN orthopedic surgeon Kuntscher (1934) was the first to investigate stress strain phenomena in human bones by using a strain sensitive lacquer He coated long bones with melted colophonium and studied the strain patterns produced by static and dynamic loading of the bone in various positions The principles of the method were discussed in Chapter Two

With static vertical loading of the femur (Figure 16a) the first cracks appeared in the colophonium on the superior aspect of the neck With continued loading a second group of cracks arose on the lateral aspect of the shaft just inferior to the level of the lesser trochanter The cracks on the neck were transverse to the long axis of the region but with increasing load they extended onto the side of the neck and curved in the direction of the shaft The cracks on the shaft were at first transverse to the long axis of the bone but later extended onto the anterior and posterior aspects where they curved inferiorly In each region the cracks arising later were parallel to those formed previously All the cracks arose from tensile strain in the underlying bone and were transverse to the direction of the strain The superior *aspect of the neck was the area of highest tensile strain since the* first cracks appeared there

Areas of compressive strain produced by static loading of the femoral head (Figure 16b) were also investigated In these tests cracks appeared in the colophonium as the load was gradually removed, thus permitting the region under compression to return to its original condition During the return tensile strain in the bone gave rise to cracks in the overlying colophonium Consequently the cracks indicated areas of the bone which had been subjected to compressive strain during a test The area of highest

pressure point on the lateral condyle but in a bone having a well developed genu valgum the medial condyle showed the greater compressive stress In a tibia from a person with chondrodystrophy application of pressure to the joint surfaces i e, at opposite ends of the bone produced a band of tensile strain extending across the shaft from its most convex point on the lateral aspect to its most concave point on the medial aspect of the bone The pattern indicated the presence of tensile forces in the living bone which resulted in the formation of resorption (Looser s) zones In a markedly bent tibia the points of compressive strain were proximal and distal to the area of bending The deformation pattern produced in a rachitic femur showed that the direction of the tensile strain was approximately in the long axis of the bone similar to that of the split lines Benninghoff obtained on the other femur from the same individual

In torsion tests of deformed bones the tensile strain did not spiral around the shaft as in normal bones but was restricted to either side of the bent area (Figure 16d) In normal bones the cracks constituting a torsion pattern had a 45 angle to the long axis of the shaft but this could be changed by additional tensile or bending stress Normally the peak of the tensile strain was in the middle of the shaft Kuntscher stated that the greatest strain was produced by torsion although he did not record the magnitude of the strain or of the load producing it

Kuntscher believed that deformations occurring in mature bones arise from tensile stresses while those in growing bones are produced by pressure stresses in the growth area Certain pseudoarthrodeses and congenital coxa vara arise, according to him in areas of high tensile strain where resorption areas (Looser s Zones) develop

From the uniformity of the deformation patterns obtained in his tests Kuntscher concluded that an intact bone, in spite of its anatomical diversity is a functionally homogeneous structure A similar conclusion had previously been reached by Messerer (1880) Kuntscher stated that the strain patterns obtained in his tests were experimental confirmation of Benninghoff's concept (1925) that the osteones (Haversian systems) are functionally related to the stresses and strains to which the bone is subjected

compressive strain was the inferior aspect of the femoral neck. With heavier loads the strain pattern spread onto the medial aspect of the shaft. The cracks themselves were perpendicular to the long axis of the bone and the direction of the strain. Areas of tensile and compressive strain within a single bone were also demonstrated (Figure 16b).

Kuntscher pointed out that bending and torsion are the types of force most frequently applied to the long bones in the living body. In torsion tests of the tibia he found that the resulting strain pattern consisting of obliquely placed cracks spiraled around the shaft of the bone at a 45° angle to its long axis (Figure 16c).

In another series of tests Kuntscher (1935a) determined the areas of tensile and compressive strain produced by tension, compression, bending, and torsion loading of a normal humerus, tibia, femur, two radii, and a clavicle. The areas of strain varied with the bone, its orientation during a test, and the type and point of application of the force.

Kuntscher then published a paper (1935b) in which he gave additional data on static loading and extended his studies to include dynamic loading of pathological bones and a discussion of fracture mechanisms. Dynamic loading was obtained by dropping weights upon the head of the feumr or striking it with a hammer. The magnitude of the energy thus applied to the bone was not stated.

The tensile strain patterns produced by dynamic loading of normal femurs were essentially similar to those obtained by static loading. Tilting the bone so that the shaft made a different angle with the horizontal plane had little effect upon the deformation pattern. Among the normal bones tested were femurs from two girls 14 and 17 years of age and one from a newborn boy. Cracks were found on the superior aspect of the neck distal to the epiphyseal line. Kuntscher interpreted these findings as evidence that the earliest mechanical influence on the region of the epiphyseal line is tension on the superior part of it.

The deformation patterns obtained on deformed bones were quite different from those of normal bones. Thus the lower end of a normal femur subjected to static loading had the higher

pressure point on the lateral condyle, but in a bone having a well developed genu valgum the medial condyle showed the greater compressive stress In a tibia from a person with chondrodystrophy application of pressure to the joint surfaces i e, at opposite ends of the bone, produced a band of tensile strain extending across the shaft from its most convex point on the lateral aspect to its most concave point on the medial aspect of the bone The pattern indicated the presence of tensile forces in the living bone which resulted in the formation of resorption (Looser s) zones In a markedly bent tibia the points of compressive strain were proximal and distal to the area of bending The deformation pattern produced in a rachitic femur showed that the direction of the tensile strain was approximately in the long axis of the bone similar to that of the split lines Benninghoff obtained on the other femur from the same individual

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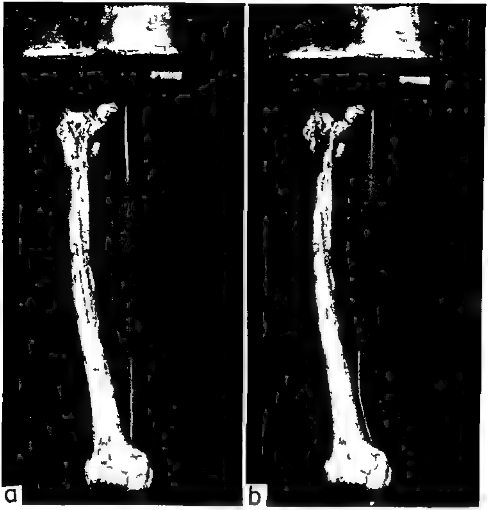


Figure 17 Human femur in a testing machine a—unloaded bone b—same bone under a 650 lb load (From the original of Figure 7 Evans and Lissner *Anat Rec 100* 175 1948)

In addition he believed that the osteone lines of the proximal end of the femur have a double function in that they transmit both tensile and compressive forces to the superior aspect of the neck However on the inferior aspect of the neck they resist compression forces only In Kuntscher s opinion roentgenograms of the femoral neck confirmed Benninghoff's idea that the trabeculae represent continuation of the lines of the osteones In addition he stated that the trabeculae are directly related to the direction of tensile strain in the femoral neck The latter idea

conflicts with Kuntscher's criticism of Culmann's interpretation of the trabeculae as representing trajectories In attacking Culmann's analysis of the proximal end of the femur Kuntscher emphasized that the trabeculae only resemble the maximum internal stress trajectories in a crane when the femur is sectioned in the frontal (coronal) plane, and that a section in another plane presents an entirely different picture A similar criticism might apply to Kuntscher's roentgenograms which show only an anteroposterior view of the femur

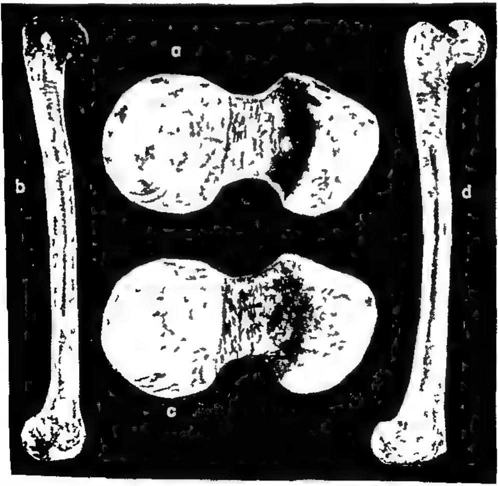


Figure 18 Stresscoat patterns produced in the femur of a white male 60 years of age by a static vertical load of 480 lbs (a b) and a dynamic vertical load of 15 8 in lbs of energy (c d) (From the original of Figures 12 12a 13 and 13a Evans Lissner and Pedersen *Anat Rec* *101* 239 1948)

Results With the Stresscoat Method

Recently the distribution of tensile strain in the human femur under various conditions of loading and orientation has been restudied by means of stresscoat an industrial method more sensitive and easier to use than colophonium The principles of the method are fully discussed in Chapter Two

The first stresscoat studies on long bones were those of Evans and Lissner (1946), who analyzed the deformation patterns produced in normal adult human femurs by static vertical loading The femurs were obtained from embalmed cadavers, whose age sex, race, and cause of death were known and were tested in a Baldwin Southwark materials testing machine calibrated to an accuracy of ± 0 5% The load was statically applied to the head of the vertically oriented bone (Figure 17a) This orientation was chosen because according to Walmsley (1932) it closely approximates the positon of the femurs when standing erect with the heels together

During a test the femur behaved like an eccentrically loaded column subjected to a bending action (Figure 17b) The bending creates tensile strain on the convex aspect of the bent bone as shown by the appearance of a deformation pattern on the superior aspect of the neck and the lateral aspect of the shaft (Figure 18a and b) The first stresscoat cracks, indicating the site of highest tensile strain appeared on the superior aspect of the neck just distal to the head Slightly later other cracks arose on the shaft With increasing load additional cracks appeared in both areas In bones with marked anterior bowing the deformation pattern gradually spread from the lateral aspect of the proximal third onto the anterior aspect of the middle and distal thirds of the shaft

The tests showed that the intact femur when subjected to nonfracturing loads acts like an elastic body returning to its original shape and *dimensions when the load is removed In* addition it was found that under a static vertical load the lateral condyle supports a greater proportion of the load than does the medial one

Two series of tests with lacquers of different sensitivity were

made In the first test series, with a lacquer sensitivity of 0 0018 inches/inch the minimal load necessary to produce a tensile strain pattern on the superior aspect of the femoral neck varied from 560 pounds in the right femur of a white male 60 years of age, to 990 pounds in the right femur of a white man of unknown age In the second test series, with a lacquer sensitivity of 0 0012 inches/inch the load necessary to produce a threshold pattern on the neck ranged from 400 pounds for the left femur of a Negro male 49 years of age, to 700 pounds for the corresponding bone from a white male 57 years of age The average load applied in the first and second series of tests was 720 pounds and 646 pounds, respectively In all the bones tested the orientation of the stresscoat cracks indicated that static vertical loading of the femoral head creates tensile strain in the direction of the long axis of the neck and the shaft of the bone

An attempt was made to correlate the deformation patterns with the age of the individual from whom the femur was obtained and such anatomical features as the weight and maximum length of the femur the diameter of the neck and the shaft and the magnitude of the angle which the vertical the neck and the shaft axes made with one another It was found that femurs from individuals less than 60 years of age required a greater load to produce a minimal deformation pattern than did bones from older individuals In addition the larger heavier bones supported a greater load before the appearance of a deformation pattern A slight tendency was noted toward a positive correlation between the shaft diameter and the load necessary to produce a minimal deformation pattern Of the various angles measured the one between the axes of the neck and the shaft was the most important with respect to the weight bearing capacity of the femur The two bones with the largest neck shaft angle supported the greatest load before the appearance of a strain pattern although no consistent relation was found between the magnitude of this angle and the load necessary to produce a minimal deformation pattern

The stresscoat deformation patterns produced by dynamic vertical loading of the human femur were likewise studied (Evans Lissner and Pedersen 1948) In these tests the vertically ori

Results With the Stresscoat Method

Recently the distribution of tensile strain in the human femur under various conditions of loading and orientation has been restudied by means of stresscoat, an industrial method more sensitive and easier to use than colophonium. The principles of the method are fully discussed in Chapter Two.

The first stresscoat studies on long bones were those of Evans and Lissner (1948), who analyzed the deformation patterns produced in normal adult human femurs by static vertical loading. The femurs were obtained from embalmed cadavers whose age, sex, race, and cause of death were known, and were tested in a Baldwin Southwark materials testing machine calibrated to an accuracy of $\pm$ 0.5%. The load was statically applied to the head of the vertically oriented bone (Figure 17a). This orientation was chosen because, according to Walmsley (1932), it closely approximates the positon of the femurs when standing erect with the heels together.

During a test the femur behaved like an eccentrically loaded column subjected to a bending action (Figure 17b). The bending creates tensile strain on the convex aspect of the bent bone as shown by the appearance of a deformation pattern on the superior aspect of the neck and the lateral aspect of the shaft (Figure 18a and b). The first stresscoat cracks, indicating the site of highest tensile strain, appeared on the superior aspect of the neck just distal to the head. Slightly later other cracks arose on the shaft. With increasing load additional cracks appeared in both areas. In bones with marked anterior bowing the deformation pattern gradually spread from the lateral aspect of the proximal third onto the anterior aspect of the middle and distal thirds of the shaft.

The tests showed that the intact femur, when subjected to nonfracturing loads, acts like an elastic body, returning to its original shape and dimensions when the load is removed. In addition, it was found that under a static vertical load the lateral condyle supports a greater proportion of the load than does the medial one.

Two series of tests with lacquers of different sensitivity were

Later experiments (Pedersen Evans and Lissner 1949) showed that tilting the femur so that the infracondylar plane made a 3 angle with the horizontal plane had little influence upon the location of the tensile strain pattern produced by dynamic vertical loading It was found that 23 7 inch pounds of energy were necessary to produce a threshold pattern when a lacquer with a sensitivity of 0 00105 inches/inch was used

In a second series of studies the right femurs were tested with 23 7 inch pounds and the left with 31 6 inch pounds of energy The stresscoat cracks on the left femurs were always closer together than those of the right bones indicating a greater deformation or strain in the former because of the greater energy applied to them

Twelve femurs were tested in a third study by dynamically applying the load to the greater trochanter such as occasionally occurs in a fall This was termed abduction" loading because, theoretically it placed the inferomedial aspect of the neck and the shaft under tensile strain as happens when the femur is forcibly abducted with the proximal end relatively fixed With a lacquer sensitivity of 0 00085 inches/inch approximately 20 inch pounds of energy were necessary to produce a tensile strain pattern

The deformation patterns obtained in these tests (Figure 19a) showed that "abduction loading created tensile strain in the inferior aspect of the neck and the adjacent medial aspect of the shaft The extent of the strain varied with the size of the bone and the magnitude of the energy applied to it In similar tests under a load statically applied in a testing machine the first stresscoat cracks appeared on the inferior aspect of the neck just distal to the head indicating that it was the area of highest tensile strain where failure should occur with sufficient load With increasing load the strain pattern extended distally along the inferior aspect of the neck and onto the medial aspect of the shaft Variations in the size of the neck shaft angle had little influence upon the deformation pattern No correlation was apparent between the age of the individual from whom the bone was obtained and the extent of the strain pattern

A well defined tensile strain pattern was frequently found on

ented femur rested upon a 160 pound steel block while a 7 9 pound brass block was dropped vertically upon the head of the bone In each test the brass block was caught by hand on the rebound so that it struck the bone just once The weight of the brass block multiplied by the distance through which it was dropped gave the energy (inch pounds) dynamically applied to the bone Because of the small magnitude of the energy involved in the tests it was assumed that all of it was spent in deforming the bone since the amount absorbed by the steel slab was negligible Measurements similar to those taken on the bones used in the static loading studies were made plus additional ones of the curvature of the bone at points ¼, ½ and ¾ the length of the shaft from the tip of the greater trochanter

Fourteen femurs from nine individuals were studied in two series of tests The sensitivity of the stresscoat lacquer in the first and second series was 0 00085 inches/inch and 0 00095 inches/inch, respectively In each case the energy applied to the bone was 15 8 inch pounds

Although the results obtained from testing the same bone under static and dynamic vertical loading are not directly comparable because the first condition involves the use of pounds and the second the use of energy it was significant that the location of the tensile strain pattern on the bones was similar in both tests (Figure 18) Furthermore the studies showed that a relatively small load suddenly applied can produce a deformation pattern similar in extent to that arising from a much greater load gradually applied In both conditions of loading the magnitude of the tensile strain at any given point on a femur is directly proportional to the perpendicular distance of that point from the load axis of the bone The distance itself depends upon the curvature of the shaft and the angles which the vertical the neck and the shaft axes make with one another

The amount of bending of the bone and hence the location and extent of the tensile strain is influenced by the diameter of the neck and of the shaft The curvature of the shaft also affects the location of the deformation pattern which is concentrated upon the middle third of the anterior aspect of the shaft in those bones with a marked anterior bowing

was assumed, absorbed a negligible amount of energy used in a test. Consequently, practically all of the energy expended in a test was absorbed by the bone.

The influence of torsion was studied by testing the femurs in a torsion machine calibrated to an accuracy of ± 0.5%. The axis about which torsion occurred passed through the center of the femoral shaft. Special clamps were made to hold the proximal and distal ends of the bone. During a test the distal end of the femur was fixed, while the head and greater trochanter were rotated as a unit in a medial direction. Thus lateral rotation of the thigh, with the proximal end of the leg fixed, was simulated. The amount of torsion (torque) applied to a femur was automatically recorded in units of 8.3 inch pounds of torque by a dial gage on the torsion machine. Twelve femurs, coated with a lacquer having a sensitivity of 0.00095 inches/inch, were tested with loads from 149.4 to 514.6 inch pounds of torque.

Torsion loading produced a tensile strain pattern (Figure 20a) which spiraled around the shaft at a 45° angle to its long axis. During a test the anterior and posterior aspect of the neck were placed under tensile strain by the torsion force which was applied to the anterior aspect of the head and the posterior aspect of the greater trochanter. Although the first cracks appeared on the neck, the clamps held the proximal end of the femur and thus prevented the tensile strain from exceeding the elastic limit of the bone and thus producing a fracture. No consistent relation was noted between the age of the individual whose bones were tested and the magnitude of the torque necessary to produce a strain pattern. The direction of the stresscoat pattern was in accord with mechanical principles that torsion produces tension along a spiral path around the center of rotation. When the axis of rotation is vertical in position the spiral path makes a 45° angle with the horizontal plane. The tensile strain is opposed by a compressive strain which spirals around the center of rotation in the opposite direction, the paths of the two types of strain crossing at a right angle.

The extent of the tensile strain patterns produced in human stresscoated femurs by application of controlled amounts of energy to target points ¼, ½, and ¾ the length of the shaft from

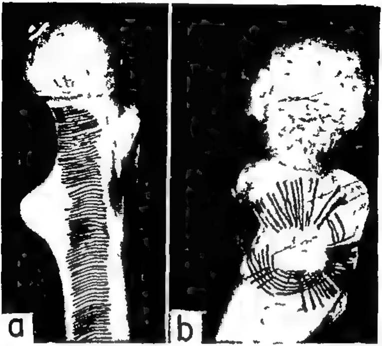


Figure 19 Stresscoat patterns produced by applying 31 6 in lbs of energy to the greater trochanter (From the original Figures 14 & 15 Pedersen Evans and Lissner *Anat Rec* 103 181 1949)

the greater trochanter at the site of impact (Figure 19b) The pattern arose from inbending of the bone by the load applied to it and when fully developed consisted of circularly and radially oriented cracks crossing one another at right angles The orientation of the cracks indicated that inbending of the bone created tensile strain in a circular and a radial direction simultaneously The circular strain produced the radially oriented cracks while the radial strain gave rise to the circularly oriented cracks

In some of the tests the femoral head was gradually elevated from 73 mm to 290 mm above the floor while 31 6 to 355 inch pounds of energy were applied to the greater trochanter Although elevating the femoral head reduced the effective bending force acting on the neck and increased the direct compressive force acting on the shaft it produced no significant differences in the tensile strain pattern In the abduction loading tests the head of the femur rested upon a 160 pound steel slab which it

Femurs from two Negro females, 16 white males, and three Negro male cadavers were tested Both femurs from 14 individuals were used but in four bodies only the right bone and in two others only the left bone were intact for testing All bodies were embalmed The age of the individuals varied from 38 to 85 with an average age of 61 1 years for the whole series

The energy was applied to the femur by dropping a 7 9 pound brass block upon a solid steel rod one inch in diameter, laid across the shaft of the bone at the target point The rod concentrated the energy at the target The brass block was caught by hand on the rebound so it struck the steel rod just once The ends of the femur were elevated upon steel blocks so that the target was horizontal in position The magnitude of the energy used in the experiments was so small it was assumed that none of it was absorbed by the steel blocks and rod

The anteroposterior and transverse diameters of the shaft, as well as its cross section area, were determined at each target point The area was computed from the formula for an ellipse πab where a is the radius of the shaft in the anteroposterior direction and b its radius in the transverse direction The curvature of the shaft at each target site was determined by measuring the projected vertical distance from the table top, upon which the femur rested on its posterior aspect to the target point on the anterior aspect of the shaft The tensile strain pattern produced in a test was measured and its extent in terms of per cent for shaft length was computed The latter served as the basis for comparison of the extent of the strain produced in the various experiments

The chief differences found in the experiments were the aspect of the femur subjected to tensile strain, the extent of the strain and the occurrence of local strain patterns at the site of impact The strain patterns, regardless of the aspect of the shaft involved indicated that tensile strain in the direction of the long axis of the shaft had been produced from the bending of the bone as a result of the energy applied to the opposite side When the energy was applied at the ¾ point of the shaft, local strain patterns frequently arose from inbending of the bone at the site of impact (Figure 20b) The sensitivity of the lacquer

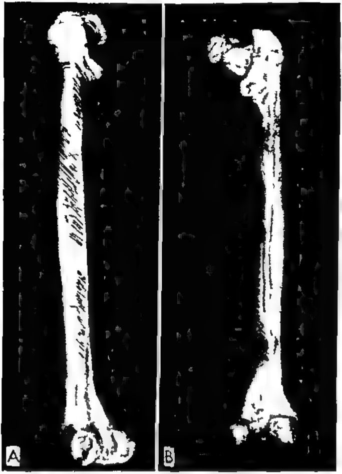


Figure 20. Stresscoat patterns produced by different types of loading. *a*—Pattern produced by 514.6 in. lbs. of torque. *b*—Pattern produced by 11.8 in. lbs. of energy applied to the ³⁄₄ point of the lateral aspect of the shaft. (*a* from Evans, Pedersen and Lissner, *J. Bone & Joint Surg.*, 33-A, 1951. *b* from the original Figure 4c, Evans, Hayes and Powers, *Anat. Rec.*, *116*:187, 1953.)

the tip of the greater trochanter has been investigated by Evans, Hayes and Powers (1953). In their experiments 7.9 inch pounds of energy were applied to the ½ point and 11.8 inch pounds to the ¼ and ¾ points of the anterior, posterior and lateral aspects of the shaft. The medial aspect of the shaft was not tested because in the living individual it is rarely subjected to a fracturing blow (Figure 20b).

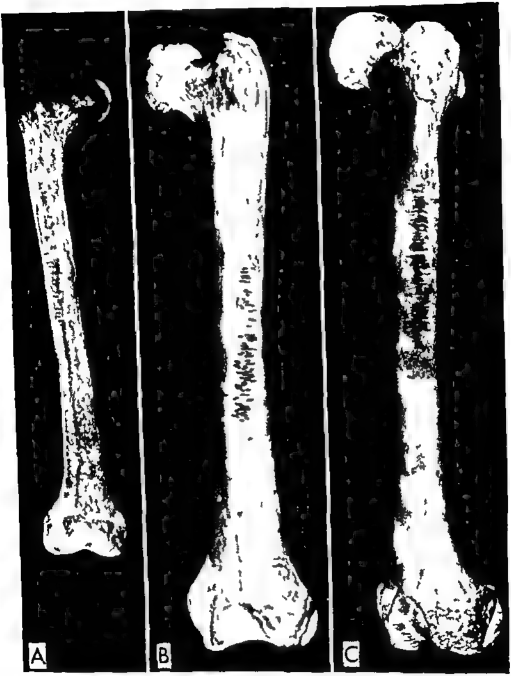


Figure 21. Stresscoat patterns produced by static vertical loading *a*—Pattern produced in the femur of a white male 61 years of age by 450 lbs *b*—Pattern produced in the femur of a female gorilla by 310 lbs *c*—Pattern produced in the femur of a male orangutan by 300 lbs

used in the experiments varied from 0 0004 to 0 0009 inches/inch

The most extensive strain in the anterior aspect of the shaft (61% of its length) arose from application of the energy at the ¼ point of the opposite side of the bone In the posterior aspect of the shaft the most extensive strain (55% of the shaft length) occurred from loading the ¾ point of the opposite side Loading the corresponding point of the medial aspect of the shaft produced the most extensive strain (63% of the shaft length) on the lateral aspect of the bone In each instance the figure represents the average extent of the strain

In the stresscoat experiments previously discussed the extent of the tensile strain pattern in the various femurs varied with the mass of the bone itself Characteristic patterns were produced for each type of loading and from them the point of application and type of force applied to the bone can be deduced Conversely if the type of force and its point of application are known the location and characteristics of the resulting strain pattern can be predicted

The same was true of the strain patterns Kuntscher obtained with the colophonium method Evans Lissner and their associates did not study deformed bones but the tensile strain patterns they obtained in normal bones closely resembled those found by Kuntscher under similar experimental conditions

Comparative studies the details of which will be published elsewhere of the tensile strain patterns produced by static vertical loading of stresscoated femurs of nonhuman primates have been made by Evans and Straus The femurs were tested in a 5000 pound capacity Riehle testing machine calibrated to an accuracy of ± 1% The resulting stresscoat patterns (Figures 21 and 22) clearly showed that the biomechanical behavior of the femurs under static vertical loading was essentially similar to that of human femurs under corresponding conditions All the strain patterns obtained in the nonhuman primate femurs could be closely duplicated in one or more of the human femurs previously tested Femurs of some Great Dane dogs were also tested and gave tensile strain patterns closely resembling those of human femurs

differences, such as size, thickness, etc., of the bone. It is generally held that the shape of the human femur is specifically adapted for upright posture but the results from the comparative stresscoat studies do not substantiate this belief. Rather, the resemblance of the strain patterns on the femurs of pronograde mammals (dog, macaque) to those of human femurs indicate that the former are equally well adapted for supporting a vertical load. This is not surprising in view of the similarity in shape of most mammalian femurs.

Results With Electric Strain Gages

Evans, Coolbaugh, and Lebow have used electric strain gages, of the SR4 type, to record stresses and strains occurring in the tibia of a living dog. The gage was cemented on the medial aspect of the tibia and the tensile and compressive strains (Figure 23) occurring in the bone while the dog was walking about were recorded on a cathode ray oscilloscope. Tensile strain was indicated by the spikes above the base line and compressive strain by those below it.

Each type of strain arose from bending of the tibia from the weight of the dog. By placing additional gages on other aspects

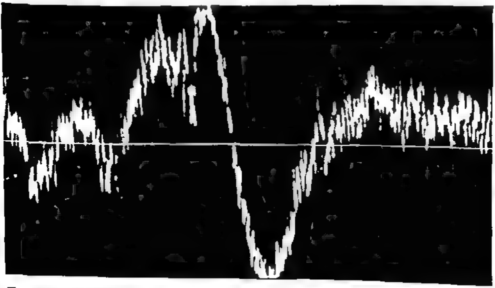


Figure 23. Oscilloscopic record of strains in tibia of a walking dog. (From the original Figure 15, Evans, *Am. J. Phys. Anthropol.*, *11*:435, 1953.)

Figure 22 Stresscoat pattern produced by static vertical loading *a*—Pattern produced in the femur of a Negro male 49 years of age by 520 lbs *b* and *d*—Pattern produced in the femur of a Rhesus monkey by 50 lbs *c*—Pattern produced in a human femur by 1280 lbs

In all femurs regardless of the postural habits of the animal from which the bone was obtained the superior aspect of the neck and the lateral aspect of the shaft were the regions subjected to tensile strain However the extent of the strain was modified by the magnitude of the load and individual anatomical

7

The Magnitude of Stress and Strain in Long Bones

THE MAGNITUDE of tensile and compressive stresses and strains in long bones has been measured by several investigators Intact bones were generally used

Measurements With Extensometers

The tensile and compressive strains produced by static vertical loading of adult human femurs were measured by Kuntscher (1936) with Okhuizen extensometers having a gage length of 10 to 20 mm and an accuracy of ± 0 1% Most of the bones were tested two or three days after removal from the body and were unembalmed

During a test the femur was held vertically in place by fixing the distal end in a metal can The load was statically applied to the femoral head through a lever calibrated in centimeters to which a weight was attached The magnitude of the load applied to the bone was varied by sliding the weight along the lever arm A rubber pad was placed between the lever and the bone so that the load was evenly distributed over the femoral head

The anterior aspect of the proximal two thirds of the femur was divided into six vertical and 19 horizontal coordinates 1 cm apart The first vertical coordinate passed through the tip of the greater trochanter and the sixth touched the most medial point of the femoral head The first horizonal coordinate passed through the femoral head just anterior to the foeva capitis and the greater trochanter 1 cm distal to its tip The nineteenth horizontal coordinate passed through the bone at the approximate junction of the middle and distal third of the shaft

Four measurements over a distance of 10 to 20 mm, were taken at each intersection of the coordinates The planes of the

of the tibia it would have been possible to record the tensile and compressive strain produced by bending of the bone in other planes In addition the gage and oscilloscope could have been calibrated so that the actual magnitude of the various strains could be measured

Summary

The results of the studies with strain sensitive lacquers show that the long bones behave like elastic bodies, which can be deformed with relatively small amounts of energy or rather light loads

When the load is applied to the head of the vertically oriented femur the superior aspect of the neck and the lateral aspect of the shaft are subjected to tensile strain while the opposite aspects of the bone are under compressive strain. Both types of strain arise from bending of the bone

Loading the shaft of the femur perpendicular to its long axis also causes bending of the bone with tensile strain created on the convex aspect of the bone opposite that to which the load was applied The latter aspect especially under the point of application of the load is subjected to compressive strain

Loading the greater trochanter of the femur creates tensile strain in the inferior aspect of the neck and the immediately adjacent area of the medial aspect of the shaft Local deformation patterns produced by inbending of the bone frequently occur at the site of impact or loading They arise from tensile strain created by the pushing in of the bone at the loading site

The strain sensitive lacquers show the over all distribution of strain in a bone From the deformation patterns obtained the sites for placing strain gages of various types in order to measure accurately the magnitude of the strains are easily determined

The data show that the deformation in bone is directly proportional to load and that the femur, in spite of its structural diversity follows Hooks Law as Kuntscher stated

The tensile and compressive strains produced in adult femurs by static vertical loading in an Amsler Testing Machine were measured by Marique (1945) with Huggenberger extensometers having an accuracy of approximately ± 1% Two extensometers to serve as controls, were placed opposite one another in the center of the shaft, while two other pairs were placed at various levels above and below the control pair Most of the measurements were taken over a gage length of 10 mm and gave the longitudinal strains occurring along the anterior and posterior aspects and the external and internal borders of the shaft of the bone However the measurements did not necessarily represent the maximum and minimum values for the strains at different levels of the femoral shaft

Before testing each femur was weighed and its length measured Roentgenograms were taken of each bone from which the characteristics of the trabeculae of the head the thickness of the shaft compacta the neck shaft and declination angles and the magnitude of the inclination of the condylar end of the bone to the shaft and to the mechanical axes were determined

Two right (I and II) and one left (III) femurs were tested under loads of 50 150 and 250 kgs The bones were defleshed and the right ones tested in an unembalmed moist condition The left femur was a dry museum specimen One of the right bones (II) was from a white male 42 years of age who died of pneumonia the day before the tests No data were given for the other bones During the study 538 readings from 12 consecutive mountings of the extensometers were taken on the first femur 429 readings from 14 mountings on the second and 318 readings from 12 mountings on the third

The maximum tensile (+) and compressive (−) strains as well as their site in the bone produced by loads of 100 kg and 200 kg are given in Table I The magnitude of the strain was measured in thousandths of a mm for a gage length of 10 mm The distance of the site of maximum strain from the distal end of the bone was recorded in centimeters

measurements each separated from the adjacent one by a 45 angle, are indicated in the accompanying diagram

The magnitude of the strain was recorded in thousandths of a mm and the measurements were made under a 200 kg load vertically applied to the femoral head

The maximum tensile strain (0 0065 mm/mm) occurred in the superior aspect of the neck close to the greater trochanter The direction of the strain was parallel with the long axis of the neck The high point of tensile strain in the shaft (0 00415 mm/mm) was on its lateral aspect, about 10 cm from the tip of the greater trochanter at the intersection of the first vertical and the tenth horizontal coordinates The highest compressive strain (0 0055 mm/mm) was found on the medial aspect of the shaft at its intersection with the fifth coordinate The point of this intersection was approximately 1 cm proximal to the lesser trochanter The direction of the tensile and compressive strains in the shaft was parallel with its long axis

The magnitude of the strain was dependent upon the size of the load as evidenced from the following results obtained with different loads

Strain Site	*100 kg*	*200 kg*	*400 kg*	*800 kg*
Highest Compressive Site	0 0027 mm/mm	0 0055 mm/mm	0 011 mm/mm	0 022 mm/mm
Highest Tensile Site—Neck	0 0003 mm/mm	0 0065 mm/mm	0 013 mm/mm	0 026 mm/mm
Second Highest Tensile Site—Shaft	0 002 mm/mm	0 0041 mm/mm	0 0081 mm/mm	0 0162 mm/mm

Table I shows that the magnitude of the maximum strain produced in any of the bones was small. In all three femurs the highest tensile strain occurred in the superior aspect of the anatomical neck. The highest compressive strain in femurs I and II was in the internal border at the level of the lesser trochanter and the surgical neck respectively. In femur III the inferior aspect of the neck exhibited the highest compressive strain. This femur had a well developed coxa vara, with a neck shaft angle of 100°, which probably accounts for the increased compressive strain in the neck. Although femur III was tested dry the magnitude of the strains produced in it showed no consistent differences from those in the other two bones which were tested in a moist condition.

The elasticity of the intact femur was investigated by comparison of the ratio between the magnitude of the strain produced by different loads. The average ratios, based on 538 readings of the extensometers for femur I, 429 readings for femur II and 318 readings for femur III obtained for 100 kg. and 200 kg. loads were as follows:

	Femur I	*Femur II*	*Femur III*
Internal Border	1.92	1.90	1.90
External Border	1.99	2.21	2.09
Anterior Aspect	1.98	1.94	2.08
Posterior Aspect	1.99	1.99	2.00
General Average	1.97	2.01	2.01

The elasticity data show that in general the intact femur conforms with Hook's Law that strain is proportional to load. Femur III tested dry exhibited slightly greater strain in proportion to load than did the other two bones. However, this may have been the result of the coxa vara which made the bone less well adapted for supporting a vertically applied load.

The total tensile (+) and compressive (−) strain, in millimeters for a gage length of 10 mm., produced in the external and internal borders of the femur by a 200 kg. load was:

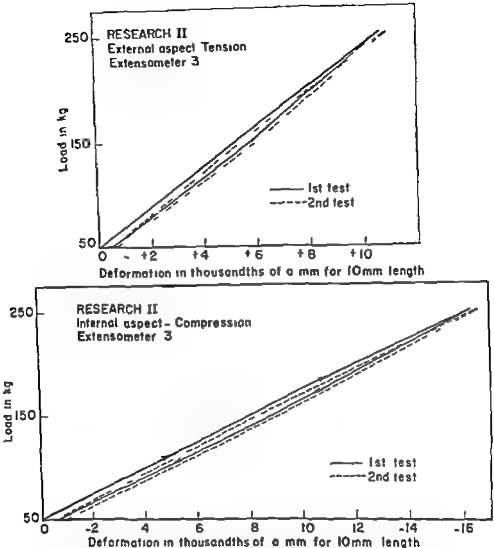
	Femur I		*Femur II*		*Femur III*	
External Border	−0.0613	+0.0654	−0.0757	+0.1346	−0.0407	+0.1734
Internal Border	−0.1537	+0.0009	−0.2332	+0.0055	−0.4399	

TABLE I

MAXIMUM TENSILE (+) AND COMPRESSIVE (−) FEMORAL STRAIN ACCORDING TO MARIQUE (1945)

	Femur I				Femur II				Femur III			
	100 kg		*200 kg*		*100 kg*		*200 kg*		*100 kg*		*200 kg*	
	Strain (mm mm)	Site	*Strain (mm mm)*	Site	*Strain (mm mm)*	Site*	*Strain (mm mm)*	Site*	*Strain (mm mm)*	Site*	*Strain (mm mm)*	Site*
Internal Border	−0.0077	lesser troch	−0.0131	below lesser troch	−0.0096	surg neck	−0.0182	surg neck	−0.0170	infer aspect neck	−0.0325	infer aspect neck
	+0.002	9–10	+0.0001	9–10	+0.009	6–7	+0.0011	6–7				
External Border	−0.0036	9	−0.0075	9	−0.0048	Int surf ext condy	−0.0175	Int surf ext condy	−0.0016	8	−0.0097	9
	+0.0043	28	+0.0081	28								
Anterior Aspect	−0.0034	30	−0.0069	30					−0.0017	27	−0.0036	27
					+0.0039	18–19	+0.0078	18–19	+0.0012	13	+0.0001	13
Posterior Aspect	−0.0024	15	−0.0050	11	−0.0061	14–15	−0.0126	14–15	−0.0124	13	−0.0238	13
	+0.0020	30	+0.0041	30					+0.0030	sup aspect neck	+0.00[illegible]	sup aspect neck
Superior Aspect of Neck	+0.0046		+0.0098		+0.0089		+0.0180		+0.0087		+0.0174	

* Distance in centimeters from distal end of bone



LOAD DEFLECTION CURVES PRODUCED BY LOADING AND UNLOADING TESTS ON FEMUR

Figure 24 Redrawn from Marque 1945

The load deflection curves (Figure 24) for increasing and decreasing loads were not exactly similar nor superimposed over one another. This phenomenon is called hysteresis. The area between the curves for increasing and decreasing loads represents the energy lost during the cycle of deformation.

Marque stated that the defect of proportionality between load and strain was influenced by aging of the bone. However, it is difficult to determine whether he meant "aging with ref

In the external border of the femur the tensile strain exceeded the compressive, just the reverse of the condition in the internal border of the bone Femur III exhibited the highest compressive strain and was the only bone in which the entire internal border was subjected to compression The latter was probably the re sult of its coxa vara In all bones the tensile strain in the external aspect was restricted to the proximal half or two thirds (Femur III) while the rest of the region was subjected to compressive strain Except in Femur I in which the two types of strain were approximately equal the tensile strain in the external aspect of the shaft was three fourths to more than three times greater than the compressive strain in the same region The compressive strain in the internal aspect of the bone was three fourths to more than ten times greater than the corresponding strain in the external aspect of the bone, and more than twice as great as the tensile strain in the same region

Additional investigations of the stress strain relationships were made by measuring over a 10 mm gage length the strains pro duced by increasing and decreasing loads of 50 150 250 150, and 50 kg The measurements were taken on the external and in ternal borders of the femur Each bone was loaded and unloaded twice The results of the tests showed that the relation between the stress and the strain at the various loads was quite similar, indicating the essentially straight line character of the stress strain curve However Marique did not illustrate any actual stress strain curves

When the load was increasing the external border of the femur was undergoing tensile and the internal border compres sive strain During decreasing load the bone tended to return to its original condition The ratio for both tests was the same for the internal aspect of Femur I *during increasing* load No bone showed complete adherence to Hook s Law although it was closely approximated for increasing load The absence of proper proportionality between load and strain during decreasing load indicated that the femur did not completely or at least im mediately return to its pre test form In only one instance the external border of Femur II during increasing load was perfect agreement with Hook s Law obtained

they obtained in their first tests varied from 0.0004 inches/inch to 0.0009 inches/inch. In six of their nine series of tests the magnitude of the strain was 0.0006 - 0.0007 inches/inch.

Assuming that stress is proportional to strain their results, based on the extent of the strain patterns indicated that the magnitude of the apparent tensile stress produced by their tests was less in femurs from individuals less than 60 years of age than in those from older persons.

Summary

The studies just discussed show that the femur behaves like an elastic body which can be deformed by various types of loading. The magnitude of the local deformation (strain) is slight and is proportional to the load. Static vertical loading of the femoral head creates the highest tensile strain in the superior aspect of the neck of the bone. Under similar conditions the greatest compressive strain is in the medial border of the bone at the level of the lesser trochanter or the surgical neck. Load deflection curves for the femur under increasing and decreasing load are not exactly the same nor superimposed over one another, indicating that the energy relations under loading and unloading are not equal.

The apparent tensile stress created in the femur by application of rather small amounts of energy (7.9 and 11.8 in. lbs.) is rather high, the greatest arising from application of energy to the 1/4 point of the lateral aspect of the shaft. Application of energy to the mid point of the posterior aspect of the shaft creates minimal tensile stress in the opposite aspect of the bone. Loading the mid point of any aspect of the bone and the 3/4 point of the lateral aspect creates tensile stress throughout the entire extent of the opposite side of the bone. The magnitude of such stress is greatest in the proximal third of the anterior aspect, the middle third of the posterior aspect, and the middle and distal thirds of the medial aspect of the shaft.

erence to the length of time he had the specimen or the age of the individual from whom the femur was obtained It could scarcely be the latter, because the age of the individuals whose femurs were tested was only stated for Femur II In any event no final conclusions on the influence of age or sex on the mechanical behavior of the femur can be drawn from just three specimens

Measurements With Stresscoat Lacquer

In connection with their studies of the extent of the tensile strain patterns produced by application of controlled amounts of energy to stresscoated femurs Evans, Hayes and Powers (*loc cit*) also investigated regional differences in the apparent tensile stress produced in the bones Energy was applied to the anterior posterior, and lateral aspects of the shaft at points ¼ ½ and ¾ the length of the shaft from the tip of the greater trochanter During the tests 7 9 in lbs of energy was applied at the ½ point and 11 8 in lbs at the other points The magnitude of the apparent tensile stress (lbs/in^2) produced in the bones was calculated by multiplying the tensile strain indicated by the value for the sensitivity of the stresscoat lacquer used by the modulus of elasticity of compact bone The figures used for the modulus of elasticity for different regions of the femoral shaft were taken from Evans and Lebow (1951) who found that the modulus was not constant throughout the bone

The maximum (2430 lbs/in) and minimal (920 lbs/in^2) tensile stress was created by loading the ¼ point of the lateral and the ½ point of the posterior aspect of the shaft respectively Loading the ¼ point of any aspect and the ¾ point of the lateral aspect of the bone produced tensile stress throughout the extent of the opposite side of the femur Under such conditions the magnitude of the stress was greatest in the proximal third of the anterior aspect the middle third of the posterior aspect and the middle and distal thirds of the medial aspect The medial aspect of the shaft exhibited the greatest and the anterior aspect the least apparent tensile stress

The magnitude of the tensile strain indicated by the sensitivity of the stresscoat lacquer responsible for the strain patterns

dynamic loading of the greater trochanter and femoral head, arise from failure of the bone because of the tensile stresses within it. This was evident from the close correspondence of the site and direction of the fracture line with the cracks in the colophonium covering the bone. Spontaneous fractures of the femoral shaft generally occurred in the region of secondary tensile strain. This was interpreted as being the probable result of architectural disturbances in the femoral neck which prevented the development of the normally occurring primary tensile peak. Only a few of the experimentally produced fractures were illustrated.

The role of tensile stresses in the production of femoral fractures was further emphasized by the stresscoat studies of Evans, Lissner and Pedersen. Under static vertical loading Evans and Lissner (1948) found that the first stresscoat cracks appeared on the superior aspect of the femoral neck just distal to the head. This indicated the site of highest tensile strain where failure or fracture should occur with sufficient load. This prediction was verified in a femur in which a transverse fracture of the neck was produced by a load of 1280 lbs. statically applied to the head of the bone (Figure 25). The fracture line began where the first stresscoat cracks appeared and paralleled the direction of the cracks as it spread across the neck of the bone. This clearly showed that the fracture arose because the bone was unable to

Figure 25. Fracture produced by static vertical loading. (From Evans, Instr. Course Lect. *Am. Acad. Orthop. Surg.* 9, 1952.) T = tensile stress.

8

The Relation of Stress and Strain to Fracture of Long Bones

THE MOST extensive investigation of the breaking strength of intact human bones was made by Messerer (1880) who tested 500 bones from 90 cadavers of both sexes and various ages. The bones were obtained as soon as possible after death and loaded to failure in a Werdersche Testing Machine. The periosteum was retained to prevent the bones from drying out during the tests which were made at room temperature (about 25° C). Messerer mentioned that the bending strength of whole bones had been tested previously by *C. O. Weber. However, he pointed out that* these earlier results were invalid because differences in bone size had been ignored.

In his own bending tests Messerer found that the cracking or tearing of the bone fibers generally occurred on the convex (tension) side of the bone. In bones exhibiting considerable bending there was crushing on the concave (compression) side at the point of application of the load before a tearing or tension fracture occurred.

Messerer usually recorded the breaking load (in kg.) but did not state the cross section area of the bones. However, he gave the ultimate breaking strength in bending for femurs from individuals from 55 to 81 years of age. This strength varied from 1330 kg/cm² in the femur of an individual 82 years of age, to 1940 kg/cm² in the femurs of two individuals 57 and 58 years of age. Both femurs from two individuals were tested but the side of the body as well as the sex of the individual were not recorded.

The influence of tensile stress in the fracture mechanism of long bones was experimentally demonstrated by Kuntscher (1935b) who pointed out that fractures produced by static and

begin in the region where the first stresscoat cracks appeared, marking the site of highest tensile strain and gradually extended across the bone in a direction parallel to that of the stresscoat cracks This proved that the fractures were initiated from failure of the bone because of the tensile stresses within it Fractures from dynamically loading the greater trochanter with the bone in the abduction position were also produced

Subcapital fractures were produced with loads of 560 and 805 lbs and intertrochanteric fractures with loads of 900 1290 and 1365 lbs (Figure 26) Horizontal fractures of the neck were

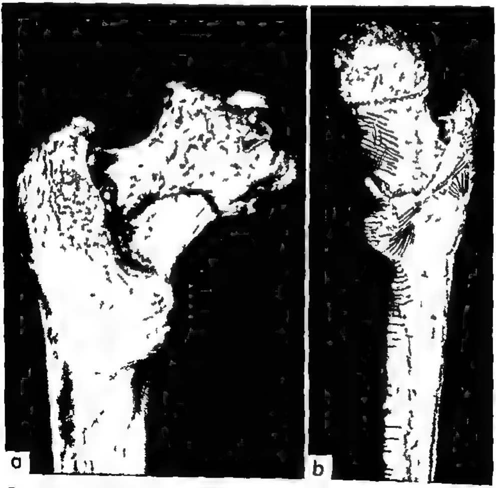


Figure 27 Fractures produced by dynamic (*a*) and static (*b*) loading of the greater trochanter (*a* from the original of Figure 17 Pedersen Evans and Lissner *Anat Rec* *103* 183 1949 *b* from Evans Pedersen and Lissner *J Bone & Joint Surg* 33 A 1951)

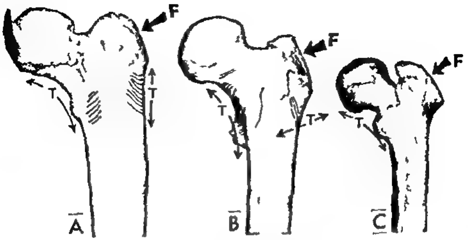


Figure 26. Fractures produced by static loading of the greater trochanter (From Evans Instr Course Lect *Am Acad Orthop Surg* 9 1952.) F = force T = tensile stress

withstand the tensile stresses created in the neck as it was bent downward by the load applied to the head of the bone

In abduction loading tests during which the femur rests upon its head and medial epicondyle static loading of the greater trochanter revealed that the first stresscoat cracks appeared on the inferior aspect of the neck a short distance distal to the head of the bone (Pedersen Evans and Lissner, 1949) In this position the femur resembles an arch whose legs are represented by the neck and shaft of the bone The inferior aspect of the neck and the adjacent medial aspect of the shaft constitute the tie beam or tension resisting member which prevents the legs of the arch from being spread apart by the load applied to the greater trochanter When the tensile stresses created in the tie beam become so great that the bone can no longer resist them the legs of the arch are pushed apart and fracture occurs

The analogy to an arch was verified by the behavior of stress coated femurs during the production of subcapital intertrochanteric oblique neck and abduction fractures from static loading of the greater trochanter in the abduction position (Evans Pedersen and Lissner 1951 Evans 1952) Each type of fracture

In some bones spiral fractures of the shaft (Figure 28a) were produced with loads of 166, 282, 148, and 498 in lbs of torque. The bones were stresscoated before testing in a torsion machine

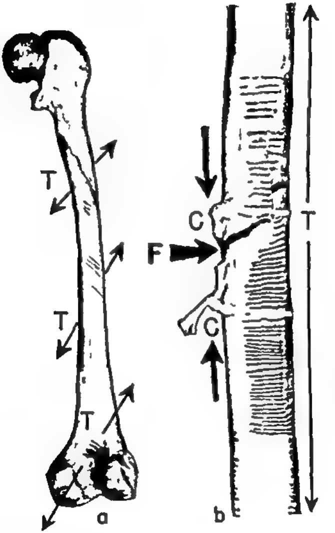


Figure 28 Two types of fracture *a*—Torsion fracture produced in the femur of a white male 60 years of age by 282.2 in lbs of torque *b*—Transverse fracture produced in the femur of a white male 79 years of age by a load of 590 lbs (From Evans Instr Course Lect *Am Acad Orthop Surg* 9 1952) C = compressive stress F = force T = tensile stress

produced by dynamic abduction loading of the greater trochanter with 308 and 344 inch lbs of energy (Figure 27a). Contrary to common belief no torsional forces were involved in producing these fractures. Torsion can only occur in an object when one end is rotated or twisted with respect to the opposite end. In the living body the femoral head and neck are entirely within the capsule of the hip joint and the only structure attached to the head is the ligamentum teres. At best, this is a weak structure and because its point of attachment to the femoral head is approximately at the center of rotation for movements of the head, it is not in a mechanical position to fix a head so that the neck can be twisted with respect to it. Therefore, in movements at the hip joint the femoral head and neck move as a unit. There is no mechanism in the body by which one can be fixed while the other is twisted.

A typical abduction type fracture (Figure 27b) which is rare clinically was produced in the right femur of a Negro female 56 years of age by 390 lbs statically applied to the greater trochanter. During the test the head of the bone was elevated three inches by resting it upon a small steel block. This fracture was unique, among those produced by abduction loading, in starting on the superior aspect of the neck at its junction with the greater trochanter. From this point the fracture was seen to extend gradually along the intertrochanteric line to the lesser trochanter. This was accompanied by an increase in the neck shaft angle until finally the neck and the shaft were firmly impacted with their axes almost in line with one another. The widening of the neck shaft angle forced the superior aspect of the neck into the greater trochanter while at the same time the tensile strain and stress on the inferior aspect of the neck was constantly increasing. Finally a second fracture started on the inferior aspect of the neck and extended across the bone to meet the first fracture line at the lesser trochanter. The second fracture arose from failure of the bone because of the tensile stresses in the inferior aspect of the neck. This particular bone originally had a neck shaft angle of 134°, considerably greater than that in the other bones and consequently was poorly adapted for withstanding a load applied to the greater trochanter.

testing machine and dynamic loading by dropping a 117 lb weight through varying distances The energy applied to the bone was computed in foot pounds No data were given about the individuals whose bones were tested or of the bones them selves except that none exhibited gross pathological conditions

Although the author states that fractures of the femoral neck are frequently the result of muscular violence, no attempt was made to determine the strength of the muscles presumably in volved or the magnitude of the forces they would exert on the femoral neck Smith found that in static vertical loading of the head of erect disarticulated femurs an average load of 2092 lbs was required to produce head intertrochanteric, subtrochanteric and neck fractures Static loading of the greater trochanter, with the femur resting upon its head and medial condyle, produced head fractures extending into the neck and intertrochanteric fractures with an average load of 1648 lbs

In articulated femurs with an intact hip joint an average load of 2149 lbs statically applied to the ilium of vertically oriented specimens produced head intertrochanteric and iliac fractures plus dislocation of the hip joint Lateral loading of the greater trochanter of similar specimens caused fractures through the ilium acetabulum symphysis pubis and sacrum with an average load of 2035 lbs Loads averaging 900 lbs applied across the neck of 20 horizontally placed disarticulated femurs resulted in transverse and oblique fractures of the neck

Smith also made similar tests with dynamic loading of the specimens In these tests it was found that an average impact of 327 foot lbs of energy applied to the head on vertically oriented disarticulated femurs produced head, neck, intertrochanteric, subtrochanteric and shaft fractures Dynamic loading of a similarly oriented specimen with an intact hip joint required an aver age impact of 250 ft lbs of energy to obtain fractures of the ilium dislocation of the hip joint and intertrochanteric fractures of the femur When applied to the greater trochanter from the side an average of 306 ft lbs of energy resulted in iliac and acetabular fractures of the pelvis and intertrochanteric and shaft fractures of the femur Subcapital oblique and transverse frac-

calibrated to an accuracy of ± 1% The spiral fractures produced by torsion loading closely followed the accompanying stresscoat deformation patterns on the bones This clearly indicates that spiral fractures arise from failure of the bone because of tensile stresses within it and not from shearing stresses as commonly believed If the latter were true the fracture line would take a transverse course across the shaft of the bone instead of spiraling around it

A transverse fracture of the shaft (Figure 28b) was produced by a load of 390 lbs applied to the middle of the shaft perpendicular to its long axis Loading a bone in this manner produced bending of the shaft The fracture began on the convex (tensile) aspect of the shaft at the point which the stresscoat pattern indicated was the area of highest tensile strain Here again the fracture arises from failure of the bone because of the tensile stresses within it Clinical examples of the various types of fractures experimentally produced were easily found in the x ray files of the Detroit Receiving Hospital

Spears and Owen (1949) used the stresscoat technique to a slight extent in a study of the etiology of trochanteric fractures However they apparently did not understand the significance of the stresscoat patterns they obtained and thought trochanteric fractures arose from failure of the bone because of shear or compressive stress In their tests both ends of the bone were fixed while the load was dynamically applied to the lateral aspect of the greater trochanter This meant that the entire trochanteric region of the bone was subjected to a bending action as a result of the impact Consequently as their photographs of stresscoat deformation patterns show the medial aspect of the trochanteric region was subjected to tensile stress and strain Very probably the fractures they discussed arose from failure of the bone because of tensile stress within it

Recently Smith (1953) has reinvestigated the magnitude of the load necessary to produce fractures in the proximal end of the human femur Disarticulated and articulated bones with the hip joint intact were tested by static and dynamic loading of the specimen in different orientations Cadaver material was used and 115 bones were tested Static loading was done in a

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amounts of tensile and compressive force fails under tension. The tensile strain patterns also enable one to predict with a high degree of accuracy where failure should occur in a bone under known conditions of loading.

tures of the neck arose from an average impact of 36.4 ft. lbs. of energy applied to the neck of a horizontally placed femur.

The type and magnitude of the stresses and strains produced in the specimens by the tests were not determined. However, in diagrams of the tests and the fractures produced, an arc of tissue tension, tending to cause tissue separation, was indicated. This seems to imply that Smith believed tension was a causative factor in the production of the various types of femoral fractures, although he did not state it in the text.

The femoral neck of 20 bones was said to have been subjected to torque, although judging from the indicated figures they were actually tested under bending instead of torsion. Smith stated that "*these torques simulated the intrinsic forces resulting* from muscular contraction of the external rotators of the hip which produce an anterior bending moment across the femoral neck." However, a torque would produce a twisting, not a bending, moment in the neck of the femur.

Smith apparently accepts the trajectorial theory of the architecture of spongy bone and shows anteroposterior roentgenograms of the ilium and articulated femur. The pelvic arch, seen in the frontal plane, was compared to a parabolic arch and one of its trabecular systems was considered as continuing into the femur where it extended from the head to the inferior aspect of the femoral neck. This trabecular system was considered as the springer of the pelvic arch.

Summary

All the various types of femoral fractures seen clinically can be experimentally duplicated under controlled conditions of loading and orientation of the bone. The close correspondence between the origin and the direction of the fracture line and deformation patterns obtained with strain sensitive lacquers proves that linear fractures of the femur arise from failure of the bone because of the tensile stresses within it. This is true of the various types of fractures of the neck and the intertrochanteric region as well as spiral fractures of the shaft of the bone.

Rauber (1876) showed that bone is weaker in tension than in compression, and consequently when subjected to increasing

amounts of tensile and compressive force, fails under tension The tensile strain patterns also enable one to predict with a high degree of accuracy where failure should occur in a bone under known conditions of loading

9

Stress and Strain in the Pelvis, Mandible and Vertebral Column

The Pelvis

Aside from the skull and the long bones stress strain phenomena have been studied most extensively in the pelvis Most of the studies were made by means of diagrams and computations of the stresses and strains produced by assumed loads However, Evans and Lissner (1955) made stresscoat studies of tensile strain distribution and fractures in actual intact pelves In the first part of their study 22 pelves, together with most of the lumbar vertebrae were tested Sixteen of the pelves were obtained from white males three from Negro males two from white females and one from a Negro female All but six of the specimens were from embalmed bodies The average age of the individuals from whom the pelves were obtained was 61 years No known pathological specimens were used

The defleshed intact pelvis including the lumbar spine was tested under dynamic loading by dropping it upon a 160 lb steel block in such a way that it landed simultaneously upon both ischial tuberosities The pelvis was released by burning a cord by which it was suspended and was caught by hand on the rebound so that the pelvis struck the steel slab just once The weight of the specimen multiplied by the distance through which it was dropped gave the energy in inch pounds dynamically applied to the specimen The method was chosen because it approximates the way in which force is applied to a pilot during emergency escape from an airplane Male pelves were tested with an average energy of 79.7 inch lbs (47 - 112.5) and female ones with an average of 39.4 inch lbs (33.1 - 45.7) The amount of energy absorbed by the steel slab would be negligible It was

therefore assumed that all the energy used in a test was expended in deforming the pelvis and spine

The tensile strain pattern produced in a stresscoated pelvis upon application of energy was visualized by spraying the specimen with statiflux powder. As the powder is sprayed upon the specimen it is given an electrostatic charge so that it collects

Figure 29 Stresscoat patterns produced in pelves by dynamic loading of the ischial tuberosities *a*—Moderately extensive in embalmed pelvis of white male 85 in lbs of energy *b*—Extensive pattern in embalmed pelvis of white male 73 years of age 288 in lbs of energy *c*—Slight pattern in unembalmed pelvis of white male 64 years of age 72 in lbs of energy *d*—Extensive pattern in embalmed pelvis of white male 82 years of age 236 in lbs of energy

along the cracks in the stresscoat lacquer and makes them visible The cracks are then traced with India ink for photographic purposes Typical examples of some of the deformation patterns obtained are seen in Figure 29

The different tensile strain patterns produced in the pelves arose from the following displacements (Figure 30) of various parts of the pelvis (1) lateral or medial rotation of the ischial tuberosities (2) lateral displacement or bulging of the acetabular regions (3) posterior displacement of the symphysis pubis (4) medial or lateral rotations of the ala and crest of the ilia and (5) various combinations of the preceding movements

Rotations of the ischial tuberosities caused bending in the ischiopubic rami with consequent tensile stress in the long axis of the region on the convex aspect of the bent ramus Tensile strain was produced on the inner and outer aspects of the ischiopubic rami by alternate medial and lateral rotations of the ischial tuberosities Lateral bulging or displacement of the acetabula created tensile strain in the long axis of the iliopubic rami, as well as within and around the acetabula This movement was accompanied by posterior displacement of the symphysis pubis

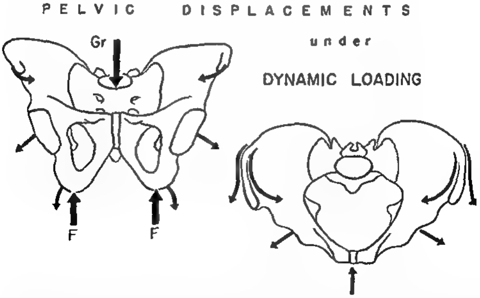

Figure 30 (From the original Figure 2 Evans and Lissner *Anat Rec* *121* 147 1955) F = force G = gravity

Rotation of the anterior superior iliac spine caused tensile strain in an anteroposterior direction, in the convex aspect of the ala The anatomical aspect of the ala involved depended upon the direction in which the rotation occurred, i e, lateral rotation created tensile strain in the medial aspect of the ala (iliac fossa) whereas medial rotation had the opposite effect In some instances a tensile strain pattern was found upon both aspects of the ala indicating that the area had undergone undulatory movement with consequent tensile strain first on one aspect and then on the opposite one

Minimal strain patterns arose when just one of the above described movements occurred, but when more than one movement was involved the tensile strain pattern was more extensive If all movements occurred the resulting pattern was very extensive, including all regions of the pelvis No relation was noted between the age of the individual from whom the pelvis was obtained and the type or extent of the deformation pattern The same was also true as far as the sex of the individual was concerned although only three female pelves were tested

In the first series of studies it was believed that too great an emphasis in the production of the deformation pattern was placed upon the mass of the pelvis itself Therefore additional studies were undertaken to determine what influence the mass of the head the trunk and the upper extremities have on deformation patterns in the pelvis Eight cadavers, from which the lower limbs the pelvic musculature, and viscera had been removed so that the pelvis could be stresscoated were studied The vertically oriented body was suspended over the steel block and dropped so that it landed simultaneously if possible upon both ischial tuberosities Care was taken to see that the ischial tuberosities were the only part of the body striking the block and that the block was struck just once The weight of the body multiplied by the distance through which it was dropped gave the energy in inch pounds dynamically applied to the ischial tuberosities The energy used varied from a minimum of 200 inch lbs to a maximum of 450 inch lbs The deformation patterns obtained in these tests (Figure 29b) were essentially similar to those produced in the intact pelvis tested outside of the body

The influence of soft tissue upon the extent and type of de formation pattern was studied in some specimens by leaving approximately ½ inch of soft tissue (mostly gluteus maximus muscle) underlying the ischial tuberosities In one such speci men a minimal deformation pattern was obtained although 450 inch lbs of energy, the most used in any of the tests, was ap plied to the specimen This indicated that soft tissue is an ex cellent energy absorbing material a conclusion previously arrived at by Gurdjian Webster and Lissner (1949) in their studies on skull deformation and fracture

The stresscoat deformation patterns produced by dynamic loading showed that the pelvis behaves like an elastic body whose deformation could be demonstrated with as little as 33 2 inch lbs of energy The presence or absence of the sacrotuberous and sacrospinous ligaments had little effect upon the type or extent of the deformation patterns produced However with similar amounts of energy the extent of the deformation patterns in the unembalmed specimens was somewhat greater than in the em balmed specimens

The deformation patterns also throw light on Braus concept (1929) of the arch like construction of the pelvis According to Braus the body weight acting through the wedge shaped sac rum subjects the pelvis to an outwardly directed tensile strain like a hat being stretched The body weight also produces stretching at the pelvic joints and tends to push apart the legs of the arch which in symmetrical standing are represented by the femurs Braus believed this outwardly directed tensile strain was resisted by counter pressure acting through the femoral heads at the acetabula Thus when standing the iliopubic rami are subjected to tensile strain but in sitting when the ischial tuberosities sup port the body weight the ischiopubic rami are the tension resist ing members

Pauwels (1948) criticized Braus concept of pelvic architec ture on the grounds that the outward push of the pelvic arch can be resisted by an inward force acting through the heads of the femurs only if the latter are firmly fixed Since the femurs are not fixed the outward push must be absorbed by the pubic rami and the symphysis pubis functioning as a tie beam or tension bar

It was also pointed out by Pauwels that the feet of the pelvic arch are represented by the supporting surface of the acetabula, not the femoral heads Consequently in symmetrical standing, the supporting surfaces are on a higher level than the tie beam, a form of arch rarely used by architects or engineers From the mechanical point of view according to Pauwels the pelvic joints cannot transmit strains of great magnitude between the bony parts of the pelvis In Pauwels analysis of the forces acting upon the pelvis in symmetrical standing the pubic symphysis is subjected to tensile strain However in the sitting position in which the effective direction of the body weight is obliquely downward and inward (from the pressure centers of the sacroiliac joints to the ischial tuberosities), the symphysis pubis is subjected to compressive strain

The deformation patterns obtained in the stresscoated pelves showed that under the conditions of the tests the iliopubic rami function as a tension bar to resist lateral displacement of the acetabula Tensile strain was also found on one or both surfaces of the ischiopubic rami However, this strain arose from bending of the rami as a result of rotations of the ischial tuberosities rather than from a more or less direct pull Generally no stress coat cracks were found immediately adjacent to the symphysis pubis which tends to confirm Pauwels contention that the symphysis is subjected to compressive strain when sitting The stress coat patterns do not agree with Benninghoff's (1925) split line patterns of the pelvis nor with his interpretation of their functional significance

The tendency for the body weight to push the sacrum between the ilia and force them apart is probably resisted by the sacroiliac and iliolumbar ligaments which thus are subjected to tensile strain This concept was verified in some pelves by the appearance of stresscoat cracks in the region of the anterior sacroiliac ligaments

Evans and Lissner also tested some pelves and lumbar spines under static vertical loading in a 5000 lb capacity materials testing machine calibrated to an accuracy of ± 1% The load was applied to the most proximal vertebra at speeds of 0 004 to 0 575 inches/minute until fracture (Figure 31a) occurred A fracture

Figure 31. Pelvic fractures produced by static (a) and dynamic (b) loading. *a*—Fracture produced in embalmed pelvis of white male 60 years of age by a load of 1350 lbs. (From the original Figure 16, Evans and Lissner, *Anat. Rec., 121*: 163, 1955.) *b*—Fracture produced in the embalmed pelvis of a white male 79 years of age by 240 in. lbs. of energy. (From the original Figure 15, Evans and Lissner, *Anat. Rec., 121*: 163, 1955.)

was also produced by dynamic loading of the pelvis in the almost intact body (Figure 31b). Comparison of the site and direction of the fracture lines with stresscoat cracks previously obtained in the same or other specimens indicated that the fracture arose from failure of the bone because of the tensile stresses and strains within it. The pelvic fractures experimentally produced are similar to those seen clinically. In addition, the stresscoat studies show that pelvic fractures are tensile failures, as previously reported for linear skull (Gurdjian, Lissner and Webster, 1947) and femoral fractures (Evans, Pedersen and Lissner, 1951).

The Mandible

Stress and strain in the adult human mandible have been studied by means of the colophonium, split line, and stresscoat techniques. Kuntscher (1934) gave a photograph of a colophonium coated mandible showing an area of high tensile strain in the approximate middle of the body of the bone and in a direction parallel with its long axis. No data were given regarding the test conditions or the magnitude of the strain.

Split line patterns of the mandible were studied by Dowgjallo (1932), who claimed they were changed by alterations in the function of the jaw following loss of teeth This was disputed by Benninghoff, cited by Murray (1936) on the grounds that some of the changes which Dowgjallo considered to be the result of functional modifications were actually within the normal range of variation in mandibles with intact dentition A modification of the split line technique was employed by Seipel (1948) in an attempt to analyze the mechanical significance of the architecture of the mandible (Figure 32) Various trajectorial systems presumably tensile and compressive were designated within the

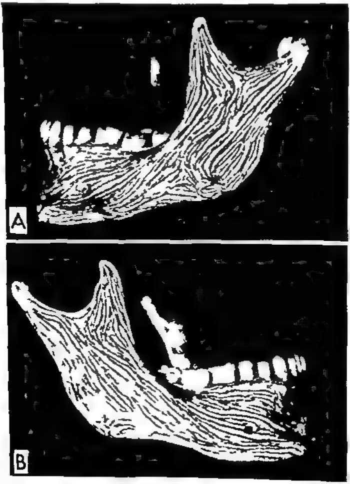

Figure 32 Split line pattern in human mandible (From Seipel *Acta odont scandinav* 8 1948)

Figure 31. Pelvic fractures produced by static (a) and dynamic (b) loading. *a*—Fracture produced in embalmed pelvis of white male 60 years of age by a load of 1350 lbs. (From the original Figure 16, Evans and Lissner, *Anat. Rec.*, *121*:163, 1955.) *b*—Fracture produced in the embalmed pelvis of a white male 79 years of age by 240 in. lbs. of energy. (From the original Figure 15, Evans and Lissner, *Anat. Rec.*, *121*:163, 1955.)

was also produced by dynamic loading of the pelvis in the almost intact body (Figure 31b). Comparison of the site and direction of the fracture lines with stresscoat cracks previously obtained in the same or other specimens indicated that the fracture arose from failure of the bone because of the tensile stresses and strains within it. The pelvic fractures experimentally produced are similar to those seen clinically. In addition, the stresscoat studies show that pelvic fractures are tensile failures, as previously reported for linear skull (Gurdjian, Lissner and Webster, 1947) and femoral fractures (Evans, Pedersen and Lissner, 1951).

The Mandible

Stress and strain in the adult human mandible have been studied by means of the colophonium, split line, and stresscoat techniques. Kuntscher (1934) gave a photograph of a colophonium coated mandible showing an area of high tensile strain in the approximate middle of the body of the bone and in a direction parallel with its long axis. No data were given regarding the test conditions or the magnitude of the strain.

parallel with the mylohyoid line as well as with the superior margin of the mandibular notch The sensitivity of the lacquer used in each of these studies was 0 0005 inches/inch In each mandible the deformation pattern arose from bending of the bone, the cracks appearing in the convex tensile areas

More recently DuBrul and Sicher (1954) used stresscoat in trying to determine the mechanical effect upon the mandible of the pull exerted by the external pterygoid muscles Stresscoated mandibles were tested by squeezing the condyles together with finger pressure applied at the points of insertion of the external pterygoid muscles However, the actual magnitude of the pressure was not stated The resulting deformation patterns (Figure 33c and d) indicated tensile strain in the long axis of the body of the mandible

The Vertebral Column

Recently considerable attention has been given to stress strain phenomena in the vertebral column especially the intervertebral discs Virgin (1951) investigated some of the physical properties of human intervertebral discs from 51 cadavers by testing them in an Olsen Testing Machine The specimens consisting of an intervertebral disc with a thin slice of bone on each end, plus sections of a lumbar spine consisting of both vertebrae and discs, were obtained at autopsy From one to five discs from each subject were tested Before and during the tests the specimens were kept in Ringer s solution The specimens were subjected to compression by successive increments of 50 lbs and the deflections occurring in the discs were measured in units of 0 0001 inches

The results were divided into two groups (1) concerning the elastic properties of intervertebral discs and (2) concerning compression of the discs In each group of studies load deflection curves were plotted

In the tests of the elastic limit of the discs the load deflection curves were not straight lines the load increasing more rapidly than the deflection The maximum load of more than 900 lbs produced a deflection of 0 0400 inches while deflections of 0 0700 inches were obtained with maximum loads of about 500 to 850

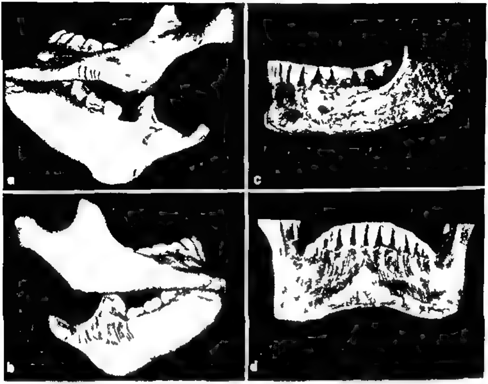

Figure 33. Stresscoat patterns in human mandible. (*b* From the original Figure 12, Evans, *Am J Phys Anthropol 11*:433, 1953; *c* and *d* from DuBrul and Sicher, *The Adaptive Chin*, Springfield, Thomas, 1954.)

mandible, although experimental proof of their functional importance was not given. Also, the magnitude of the forces assumed to be involved was not determined.

The stresscoat technique has also been used by the author in a few studies of deformations in the mandible. In one mandible a strain pattern (Figure 33a) was produced by statically applying a load of 59 lbs to the point of the chin in a direction parallel to the long axis of the body of the bone. In a second mandible (Figure 33b) a deformation pattern was produced by a maximum load of 175 lbs applied to a ¾ inch brass rod laid across the body of the mandible near its angle. In the first specimen tensile strain was produced in the long axis of the inferior margin of the mandibular body and the neck of the mandibular condyle. In the second specimen application of the load perpendicular to the long axis of the body of the mandible produced tensile strain

parallel with the mylohyoid line as well as with the superior margin of the mandibular notch The sensitivity of the lacquer used in each of these studies was 0 0005 inches/inch In each mandible the deformation pattern arose from bending of the bone, the cracks appearing in the convex tensile areas

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lbs Even if the elastic limit were exceeded the discs retained considerable power of recovery

Hysteresis was demonstrated by studies in which the loads were increased to 500 lbs and then decreased at a slower rate to 50 lbs In these tests the load deflection curve rose rapidly to a point and then fell during unloading The slope of the curve was different for loading and unloading indicating a loss in disc height of 0 005 to 0 100 inches when a load of 50 lbs was applied The magnitude of the residual deformation or strain was 0 02 inches for normal discs and 0 085 inches for immature discs

The magnitude of the residual deformation exhibited considerable regional and age variation It was smaller in the lower thoracic and upper lumbar discs and larger in the lowest lumbar discs It was very large in discs from young subjects and moderately large in those from older people in which there had been actual disc degeneration It was least in discs from middle aged subjects and undegenerated discs from people in their seventies When the disc was tested twice the magnitude of the residual deformation was always less in the second test This indicated that the mechanical efficiency of the disc had improved and that there was less energy loss during recovery

In experiments with a load of 50 lbs or less continuously applied for 48 hours, the discs showed a small deformation However they recovered rapidly after removal of the load The results of four tests on sections of whole spines were similar to those previously discussed for single intervertebral discs

Virgin s studies demonstrated that the intervertebral disc is an elastic body with considerable power of recovery Thus the strains or deformations produced in it by continual loading soon disappear after removal of the load

A somewhat more extensive study involving the mechanism of low back pain was made by Hirsch (1951) The intervertebral discs were obtained from fresh autopsied cases and were tested in a press during which not only the magnitude but the direction of the forces applied to the specimen were determined Effective vertical pressures up to 25 kg were applied by means of a lever arm The experimental devices used were such that the back could be moved in all directions or even rotated

A pressure meter containing a pelot which was thrust against the external border of a disc was designed to measure the pressure changes within a disc. The force was transmitted to the disc along a graduated scale indicating the magnitude of the pressure in kg. The distance the pelot penetrated the disc was read off on a dial indicator. Pressures of 1 and 3 kg. were used in the tests.

The specimens tested were taken from two groups of individuals: (1) those with morphologically normal discs, and (2) those with defective discs. The individuals in the two groups were about 20 and 40 years of age, respectively. Fresh autopsy material was used.

In discs from the first group the penetration of the pelot, with the same amount of pressure, was identical. However, considerable variation in the degree of penetration of the pelot was found between discs and within a single disc of the second group. Also penetration usually occurred more easily in discs in the second group.

According to Hirsch, when there is any disturbance in the structure of a disc variable pressure conditions are produced which alter the mechanical value of the disc. This alteration leads to deformations and tension disturbances which are large in some areas and which may occasionally arise very rapidly.

Additional studies were made on complete lumbar spines which were fixed in a vise so they could be turned and rotated in various directions. An apparatus consisting of sliding calipers was used to measure the exact degree of rotary movement produced in the specimen. Hirsch states that a modification in the structure of an intervertebral disc causes changes in the stress within the disc and in the path of the movement of the vertebra immediately superior to the disc. The movements of the vertebra can vary, and what takes place in the disc depends on how the forces act or the structural changes in the disc. According to him, disturbances in the mobility of the vertebra cause deformations in the intervertebral discs accompanied by stresses and strains in the longitudinal ligaments and the intervertebral joints. Hirsch considers the physical properties and characteristics of the nucleus pulposus as very important in affecting the function of the intervertebral discs. He points out that as long as the

nucleus consists of a plastic substance with a high percentage of water it contributes toward equal pressure distribution over the area of the disc. If the nucleus pulposus loses part of its plastic substance or semifluid nature mechanical stresses and strains arise in the surrounding annulus fibrosus of the disc so that it is incapable of successfully meeting the physiological demands made upon it and rupture results.

The paper was primarily a study of the effect of variations of pressure in the intervertebral discs. It was found experimentally that pressure plays a very important role in the mechanical function of the discs and herniations of the nucleus pulposus. Somewhat *similar studies were made in living individuals by injecting* salt solutions into the discs and thus altering their pressure relationships. The pressure mechanism as illustrated by Hirsch is such that the nucleus pulposus exerts pressure in all directions on the surrounding annulus fibrosus. Consequently the annulus is subjected to tensile strain in a circular direction. The effect of various movements of adjacent vertebrae on the nucleus pulposus are illustrated by diagrams. However, the actual magnitude of the various stresses and strains produced in the discs the longitudinal ligaments and the intervertebral joints was not stated.

Studies, the details of which will be published elsewhere of the magnitude of the stresses and strains produced by static vertical loading of human lumbar intervertebral discs have been made by H R Lissner and the author. The discs were taken from embalmed dissecting room cadavers because unembalmed bodies were not available. The average age of the individuals 18 white and one Negro was 57 years of age. All specimens were obtained from male individuals and consisted of the lumbar intervertebral discs and approximately half of the body of the immediately adjacent vertebrae the neural arches having been removed. The specimens were kept in embalming fluid until tested and were moist during the test.

The height of each disc plus its transverse (maximum) and anteroposterior (minimum) diameters as reflected by corresponding measurements of the adjacent vertebral body were taken in inches before and after each test. The deformations occurring in a specimen during a test were measured with a Federal

Dial gage The specimens were tested under direct compression in a 5000 lb capacity Riehle Testing Machine, calibrated to an accuracy of ± 1% The load was applied at a speed of 0.07 inches/minute until failure as indicated by a dropping off of the load on the dial of the testing machine, had occurred Deformation readings were taken at intervals of 50 lbs The load was recorded in pounds and the deformation in inches The specimen was oriented in the testing machine so that a load was uniformly applied over the surface of each adjacent half vertebra

Load deflection curves, stress strain curves, and the total energy absorbed to failure were calculated or plotted from the data obtained during the tests The approximate stresses within the discs were computed by assuming an elliptical cross section for the discs The area of the discs was obtained from the formula $A = \pi ab$, where small a and b are the major and the minor diameters The results of the test are indicated in Table II

TABLE II

AVERAGE COMPRESSIVE STRESS AND DEFORMATION IN LUMBAR INTERVERTEBRAL DISCS

Disc between	*Compressive Load (lbs)*		*Compressive Stress (psi)*		*Deformation (Inches)*	
	Average	*Range*	*Average*	*Range*	*Average*	*Range*
T12–L1 (18 spec)	801	(230–1560)	430	(164–1100)	0.0796	(0.050–0.158)
L1–L2 (15 spec)	794	(200–1495)	445	(130–1234)	0.0782	(0.043–0.100)
L2–L3 (16 spec)	925	(240–1975)	400	(145–812)	0.095	(0.048–0.148)
L3–L4 (17 spec)	935	(185–1505)	373	(116–705)	0.104	(0.054–0.197)
L4–L5 (17 spec)	885	(185–1660)	377	(122–883)	0.121	(0.044–0.280)
L5–Sacrum (13 spec)	773	(375–1240)	292	(145–850)	0.205	(0.053–0.239)

spec = number of specimen
psi = lbs/sq inch

From Table II it is seen that the disc between the first and second lumbar vertebrae exhibited the highest average compressive stress while the one between the fifth lumbar and the sacrum had the least However, the latter disc had the greatest average deformation while the former showed the least deformation

The biomechanical behavior of human lumbar intervertebral discs under both static and dynamic loading has been investigated by Hirsch and Nachemson (1954) Tests were made a few hours after death on unembalmed lumbar discs removed at autopsy No discs from cases involving tumors or tuberculosis were tested A test specimen consisted of an intervertebral disc with half of the body of the adjacent vertebrae Three tests were made on each specimen (1) with the neural arches and the vertebrae intact, (2) with a hemilaminectomy, and (3) after a total laminectomy After the tests the discs were sectioned and divided into a healthy and a degenerate group

Tests were made of 94 lumbar vertebrae, most of which were the second (94 specimens) and fourth (40 specimens) lumbar discs Five specimens were x rayed before testing The individuals whose discs were tested ranged in age from the first through the eighth decade of life the majority being between 30 and 60 years of age

Static loading studies were made in the compression testing machine used in the earlier study (Hirsch 1951) The deformation or variations in form of the discs during a test were measured by a dial indicator, having an accuracy of 0 01 mm, placed in contact with either the outer edge of a disc or the anterior margin of the body of the immediately superior vertebra Thus the actual expansion of the disc and the degree of compression were measured simultaneously Loads of 40 70 and 100 kgs were applied to the specimens

In preliminary studies it was found that the deformations shown by a given disc were reversible up to loads of 130 kgs provided the load was not applied for more than a maximum of five minutes If this load or time were exceeded the discs no longer showed recovery A slight permanent change was always present

Static loading studies of the healthy discs showed somewhat greater compressibility in the fourth than in the second lumbar disc A load of 40 kgs compressed a disc slightly less than 1 mm but a 100 kg load increased the compression to 1 4 mm It was assumed that in a person weighing from 70 to 90 kgs a 40 kg load is the approximate weight borne by the vertebral column

in the erect posture. Within the stated limitations of loading compression was proportional to stress. However, the disc must be subjected to a certain degree of compression before it exhibits a greater resistance.

Lateral expansion of the disc was somewhat less than the total compression. In 100 healthy discs, a 40 kg. load produced an anterior expansion of 0.5 mm. With a 100 kg. load the anterior expansion rose to approximately 0.75 mm. The posterior expansion of the discs was essentially the same as the anterior.

Neither hemilaminectomy nor total laminectomy produced any appreciable disturbance in the biomechanical behavior of healthy intervertebral discs. In all instances the magnitude of the deformations was the same throughout. No evidence was found to support the idea that factors other than the discs play any part in the capacity of the vertical column to support vertical loads.

Compression was more easily obtained in degenerated discs which were more sensitive to an increasing load, the compression rate rising rapidly as the load increased. A 40 kg. load produced the same expansion (0.5 mm.) as in healthy discs but with a 100 kg. load approximately 2 mm. of deformation was obtained. As in the case of a healthy disc neither hemilaminectomy nor total laminectomy had any appreciable effect on the ability of the disc to withstand a vertical load.

The authors point out that the vertebral column, like other parts of the body, is seldom in a position of static equilibrium. Consequently, static measurements have a limited value. They therefore studied the behavior of the discs under dynamic loading. In these tests the compression apparatus was *mounted on* a concrete pillar so as to eliminate, as far as possible, vibrations acting upon the machine.

Loads were dynamically applied to the specimen by letting weights fall against the lever which in turn acted upon the specimen. Weights were dropped through distances from 10 to 100 cms. with the duration of fall varying from 0.141 to 0.447 seconds. The kinetic energy from a falling weight of 1 kg. was from one to ten kilogram centimeters while that of a 2 kg. *weight varied* from two to twenty kilogram centimeters.

Before dynamically loading the intervertebral discs they were tested with weights between 10 and 130 kgs and allowed to re turn to a state of static equilibrium Blows of gradually increas ing force were then applied to the specimens All blows were vertically applied to the discs and the anterior and lateral expan sions of the discs at right angles to the direction of force, were recorded A measuring device consisting of a direct reading bridge a pre amplifier and a Kelvin and Hughes four pen re corder was used The measuring system was adapted so that when the pelot was in contact with the intervertebral disc it could be shifted in either direction and the oscillographic reading in millimeters or fractions of a millimeter directly transmitted in the path of motion The apparatus had an acuracy of $\pm$ 0 1 mm However it was found that this degree of accuracy was not abso lutely necessary so the studies were made with a sensitivity of 0 3 mm A paper speed of 5 cm/sec was used in the pen recorder although in special cases it was increased to 15 cm/sec Thus they were able to determine what was taking place within in tervals of 0 2 to 0 06 seconds The reliability of the apparatus was determined by preliminary tests on metal and wood

When a disc was subjected to a load lasting from one to two seconds compression and lateral bulging occurred However, these phenomena disappeared immediately with removal of the load indicating the disc retained practically its entire elasticity Compression experiments can be repeated indefinitely without observable disturbances in the elasticity of the discs The elas ticity curve of the discs is similar to that of normal hyaline ar ticular cartilage With greater loads the elasticity curve becomes flatter

If a disc is subjected to a violence to which it reacts rapidly it starts to oscillate or vibrate If at the moment of impact the inductive displacement pickup is allowed to absorb the move ments at any point on the surface of the disc the number of oscillations will rise in the course of 0 20 to 0 40 of a second The frequency and amplitude of the oscillations an expression of the way the discs react to the specific force vary under different conditions

Rapidly acting traumata regardless of size invariably cause

rapid oscillations which, irrespective of the strength of the blow, are always of short intensity However, with loads of 40 kgs or less the oscillations become irregular Under the same conditions the reaction of the discs is constant and can be repeated indefinitely

The results of the experiments show that the biomechanical behavior of intervertebral discs varies according to the rapidity of loading Thus if the disc is kept loaded a certain amount of compression occurs until equilibrium is obtained However, if the disc is subjected to a short momentary load it starts to oscillate In both normal and degenerated lumbar discs the magnitude of compression and lateral expansion produced by increasing loads up to 100 kgs is slight In normal discs a 100 kg load produces about 1 4 mm compression and 0 75 mm expansion In a degenerate disc with a ruptured annulus fibrosus, the same load produces a compression of 2 mm

If the load is applied for seconds perfect elasticity, of the same type as that of hyaline articular cartilage, is observed This type of compression can be repeated frequently without disturbing the mechanical response of the disc If a disc, which has assumed equilibrium under load is subjected to increasing loads of short duration it returns to its original form on unloading However the greater the load the less the shock absorbing capacity of the disc Rapidly acting loads cause an intervertebral disc to oscillate but oscillations have also been obtained with static loads of 10 to 130 kgs This shows that even after discs have assumed a state of equilibrium additional rapidly applied forces of relatively small magnitude can greatly increase the extent of the deformation Because the deformations occur frequently during relatively short periods of time (tenths of seconds) unknown stresses of considerable magnitude are produced

Great stress in the lumbar region is produced by the smallest traumata if the force is rapidly acting It seems unlikely that muscular reaction can stabilize and protect the back from the intensity and frequency of changes occurring in the shape of the disc The variations themselves are the result of the biophysical construction of the disc itself

Summary

The experiments on the intervertebral discs clearly demonstrated that the discs are very well adapted for withstanding the compressive stresses and strains to which they are normally subjected in the living body. Thus they can statically adjust themselves to various mechanical demands placed upon them. In addition, the studies of Hirsch and Nachemson show that the discs represent a dynamic system whose mass is constantly in motion. Furthermore very small rapidly applied loads or forces produce oscillations of the disc with movements measurable in tenths of a millimeter. Neither the body nor the discs are ever mechanically in a state of rest. The more frequently forces act on the discs the higher the frequency of their oscillations.

Hirsch and Nachemson believe that disturbances in the histochemical structure of the intervertebral discs are more important in the rupture mechanism than are mechanical forces. However they concede that oscillations of the frequency and amplitude found in their studies may exert an influence on the biological phenomena arising in the discs. Hirsch's earlier investigation on the mechanism of low back pain indicate that compression of a disc causes the nucleus pulposus to exert pressure on all points of the annulus fibrosus which is consequently subjected to circular tensile strain. Intervertebral discs are almost entirely compression resisting structures. Bones however are frequently subjected to bending action as a result of which they are required to resist tensile and shearing stresses and strains in addition to compression.

Stresscoat studies on the pelvis and mandible show that they also behave like elastic bodies whose deformations can be demonstrated with relatively small loads. As long as their elastic limit is not exceeded they exhibit considerable power of recovery. The importance of tensile stresses and strains in fractures was further emphasized by stresscoat patterns in experimentally produced pelvic fractures.

10

The Role of Stress and Strain in Bone Architecture

ONE OF THE first recorded comments on the mechanical significance of bone form is that of Galileo (1638) who stated that smaller bones are proportionally stronger than larger ones. During the next 200 years there were a few scattered observations on the orientation of the trabeculae in some of the long bones, but the first real attempt at a mechanical interpretation of the arrangement of trabeculae was made by Ward (1838). He studied frontal sections of the proximal fourth of the femur and compared the arrangement of the trabeculae to a triangular bracket attaching a street lamp to an upright pole. The oblique (compression) bar of the bracket was compared with the trabeculae extending from the head of the femur to the inferior wall of the neck, while the horizontal (tension) bar of the bracket was considered to be analogous to the more horizontally oriented trabeculae in the upper part of the neck.

In 1857 Wyman published a paper on the arrangement of the trabeculae in sections of the femur, vertebrae, talus, and calcaneum. He interpreted the trabecular arrangement in terms of "studs" (compression resisting bars) and "braces" (tension resisting bars) for the resistance of tension and pressure forces impinging upon the bones in the living body. His paper had originally been presented to the Boston Society of Natural History in 1849.

Humphry (1858) pointed out that in frontal sections of the femur the trabecular lines are perpendicular to the articular surface of the head and cross each other at right angles. The significance of this observation was not appreciated at the time but later it played an important role in the mathematical analysis of the functional significance of trabecular orientation.

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mur were objected to by Koch (1917) for the following reasons (1) only a small part of the femur was analyzed and its relation to the whole bone was not shown, (2) the action line of the assumed load was taken as parallel with the femoral shaft (3) the model disregarded the mass of the greater trochanter (4) although the mathematical analysis is strictly true for the model it is not applicable to the femur which has a decidedly different

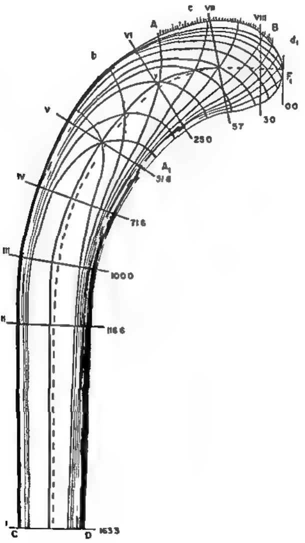

Figure 34 Culmann's trajectorial diagram of a Fairbairn crane (From Wolff *Virchow's Archiv Path Anat* 50 1870)

The next major contribution to the problem was made by Hermann von Meyer (1867) who exhibited at a scientific meeting in Zurich, sections of various human bones and discussed the functional significance of their trabecular orientation The bone sections were seen by Culmann, a mathematician and engineer, who was impressed by the resemblance between the arrangement of the trabeculae and the calculated lines of maximum internal stress (trajectories) in a mechanical structure of similar form and similarly loaded Later he computed these trajectories in a Fairbairn crane, which he assumed was a homogeneous solid structure approximating the femur in shape and compared their arrangement to that of the trabeculae as seen in a frontal section of the upper part of the femur From this Culmann concluded that the bony trabeculae were laid down along the lines of maximum internal stress thus enabling the femur to transmit a maximum load with a minimum of material

Culmann s analysis of the femur was included in von Meyer s paper and is the basis for the Trajectorial Theory of bone form According to this theory the trabeculae of spongy bone follow lines of maximum internal stress (trajectories) in the bone cross each other at right angles and arise perpendicularly from the surface of the bone or articular cartilage Certain of these trajectories are considered to be compressive and others tensile resistant

In Culmann s original trajectorial diagram of a Fairbairn crane which was the same size as an actual femur, the scale was 0 3 mm to 1 kg of force or load In making his analysis Culmann assumed that a load of 30 kg was uniformly applied over the area of the femoral head which would receive the body weight in the erect posture When Wolff (1870) reproduced this diagram (Figure 34) he reduced it to a scale of 0 15 mm to 1 kg of force The magnitude of the compressive tensile and shearing forces at eight different cross sections was also indicated The magnitude of the pressure varied from a maximum of 163 3 kg at section one to a minimum of 3 0 kg at section eight The tensile force was greatest at section one and decreased proximally, while shearing stresses absent in section one increased proximally

Culmann s conclusions regarding the architecture of the fe

with the ankylosis more complete between the medial condyles of the two bones Part of the compacta and the spongiosa had been removed before Roux obtained the specimen Although the case history of the individual from whom the specimen was obtained was unknown, Roux assumed that the concave aspect of the joint was subjected to increased pressure during walking Trajectorial diagrams of the joint were drawn on the basis of cracks produced by bending a paraffin coated rubber model of the joint

Roux's interpretation of the joint was criticized by Janssen (1920) on the basis that the actual arrangement of the trabeculae as seen in Roux s photographs of the specimen did not correspond very closely with his trajectorial diagram and that the trajectories did not behave as they should according to the Trajectorial Theory

The Trajectorial Theory of bone architecture received its best expression in 1892 when Wolff published his famous monograph entitled *The Law of Bone Transformation* According to this law every change in the form and function, or in the function alone, of a bone produces changes in accordance with mathematical laws in its trabecular architecture and external form Wolff's Law, as it has since been known was based upon the orientation of the cancellous bone in the neck of the femur

The work of von Meyer Roux and Wolff gave rise to considerable controversy regarding the factors involved in bone growth and architecture Most investigators are in agreement that function is the trophic stimulus for bone growth but there is marked disagreement as to the exact nature of the mechanical forces presumably involved Zschokke quoted by Janssen (*loc cit*) pointed out that in cases of infantile paralysis from shortly after birth the bones are essentially normal in form although they are lighter thinner and lag behind the normal bones in growth The maintenance and even growth of bones in totally paralyzed extremities was according to him proof that the form and growth of bone was of a hereditary nature entirely independent of mechanical stresses

The trajectorial nature of bone architecture was accepted by von Meyer Roux and Wolff all of whom believed that tensile

shape, and (5) the quantitative relations between the load and the area of the cross section of the femur were not determined. An additional objection is that the analysis was based upon a solid model composed of homogeneous material, whereas a femur is a hollow structure of heterogeneous composition.

Kuntscher (1934, 1935a, 1936) also criticized Culmann's analysis on the grounds that the femur is not subjected to stresses similar to those of a crane and that the model used for the analysis was simply a curved rod, not a femur. Furthermore, the trabeculae only resembled tension lines of a crane when the femur was sectioned frontally, since a section slightly away from the mid line presents an entirely different picture. Kuntscher was also not convinced of the trajectorial nature of the spongiosa.

The work of von Meyer stimulated further interest in the functional significance of the orientation of the trabeculae of cancellous bone and during the remainder of the nineteenth century numerous papers appeared on the subject. Sections at various planes were made of most of the bones of the body and the arrangement of the cancellous bone described and illustrated. In 1870 Wolff published a paper based chiefly upon the neck of the femur, in which he presented the idea that tension stresses are primarily responsible for bone growth and the orientation of the trabeculae. An opposite view was taken by Wagstaffe (1874), who studied the cancellous bone in sections taken at different planes of the vertebral column, the scapula, the pelvis, the sacrum, and all the bones of the limbs. He stated that the mechanical advantages of a bone are dependent upon its shape and surface markings, all of which increase through elasticity and strain. According to him all cancellous bone has a definite mechanical arrangement insuring the greatest strength and elasticity along the lines of greatest pressure. He pointed out that the trabecular lines seen in sections of bones are plates rather than columns and believed that pressure forces provide the trophic stimulus for the laying down of osseous tissue during bone growth.

The Trajectorial Theory was the basis for Roux's (1885) interpretation of the functional significance of the arrangement of the trabeculae in an ankylosed knee joint. The tibia of this specimen was flexed upon the femur at approximately an 60° angle

of the extensor, abductor, and lateral rotator muscles of the hip joint According to the Trajectorial Theory these two trabecular systems are pressure and tension resisting respectively

Carey attempted an experimental proof of his ideas by sectioning specific muscles in a series of dogs and studying roentgenograms of focal changes in the spongiosa of the bone He criticized the orthogonality feature of the Trajectorial Theory by pointing out many places at which the trabeculae from opposite sides of the bone did not cross one another at right angles as they should according to the theory

One of the criticisms against Carey s work is that he did not give the magnitude of the back pressure vectors, although he states that the vectors were found as the resultants of muscle forces by means of the parallelogram of forces The direction of the vectors is indicated in his figures but that is only half the story because in mechanics a vector is a quantity combining direction and magnitude Thus, both of these characteristics must be known before a vector is completely described Furthermore the direction and magnitude of the back pressure vector would constantly change during movement at a joint because the angle of application of the muscle with respect to the moving bone is constantly changing Consequently the parallelogram of force which Carey based on the muscles would also change and with it the direction and magnitude of the vector Under such changing conditions the trabeculae could scarcely represent crystallized back pressure vectors as Carey believed

One of the severest critics of the Trajectorial Theory of bone structure was Triepel who according to Murray (1936) advanced 20 objections to the theory Only the more important of these will be discussed

The first objection involves the nature of trajectories which as pointed out by Triepel are lines drawn a prescribed distance apart from a number of more or less arbitrarily selected points Theoretically the number of trajectories is infinite and if all possible ones were drawn they would arise from every possible point on the margin of the trajectorial diagram which would thus be blacked out Consequently if trajectories determine the architecture of a bone it should all be compacta without any spongiosa

stresses are primarily responsible for bone formation and growth However, this idea was denied by Janssen (*loc cit*) and by Carey (1929)

Janssen studied various types of sections of normal and pathological bones pointing out many instances in which the trabecular architecture did not conform with the Trajectorial Theory Furthermore in many cases the region of the bone presumably subjected to tensile stress had atrophied while the areas under pressure had hypertrophied In contrast to Culmann and von Meyer who interpreted femoral architecture entirely from the view of static body weight without any consideration of muscle action Janssen believed that the jerking pressure of muscles combined with that of gravity was the chief mechanical stimulus for bone formation In support of this idea he cited the vertebrae and the bones of the limbs He did however admit that other factors were involved and that not all bones e g those of the skull are the result of mechanical forces

That great force can be exerted by muscles was recognized by Christen who according to Janssen (*loc cit*) pointed out that when a man weighing 60 kg stands on tiptoe on one foot the quadriceps femoris muscle must exert a force of 240 kg in order to maintain this position It is obvious that muscles must be able to exert a force greater than the body weight in order to jump

Carey (*loc cit*) also emphasized the role of muscle action in determining the form and structure of bones Sections in different planes of the hip knee and ankle joints as well as of the entire pelvis and hallux were studied by x rays According to him the architecture of the spongy bone at mono bi and multi axial joints is entirely determined by the back pressure vectors of the muscles acting on the joints and not by static body weight For example, in frontal sections of the femur the trabecular system extending from the femoral head to the inferomedial aspect of the neck and shaft is produced by the back pressure vectors of the flexor adductor and medial rotator muscles of the hip joint The trabeculae which follow a parabolic course from the greater trochanter to the femoral head and are nearer the superior aspect of the femoral neck are produced by back pressure vectors

Triepel's idea of the close relation between the form and function of a bone is appealing but Murray (*loc cit*) points out two major objections (1) it does not agree with the facts of bone development and (2) the ideal architecture of cancellous bone, which is presumably geometrically determined, does not appear until functional demands have been placed upon the bone

Although the influence of functional stresses upon the form of bones is well established by experimental and clinical evidence other factors are involved Thus, it has been found (Murray and Huxley 1925 Murray and Selby 1930 and Murray, 1936) that in chorio allantoic grafts of limb buds from two to six day old chick embryos the humerus and the femur, in spite of complete absence of function, are often similar in form to adult bones

There have also been several studies of the effects of immobilization and absence of mechanical stresses on the gross anatomy of bones of experimental animals Allison and Brooks (1921) investigated the effects of nonuse on the foreleg bones of dogs In 13 dogs the brachial plexus was sectioned so as to produce complete or partial paralysis of one foreleg, in seven dogs the proximal end of the humerus was resected to produce a flail joint and in four dogs one foreleg was fixed in plaster of Paris The duration of the experiments before sacrifice varied from a few to 314 days

It was found that the initial changes arising from nonfunction were the same regardless of the age of the individuals However the effects of disuse in adult dogs were not the same as those in growing dogs In the former bone atrophy alone occurred but *in the latter growth was inhibited although not stopped* The experiments on growing dogs lasted from ten to 200 days while adult dogs were used in experiments of 200 to 314 days duration

In the atrophied bones the periosteum was very closely adherent to the shaft and when stripped off left a surface with the texture of fine sandpaper The thickness of the cortex of the shaft decreased with consequent enlargement of the medullary cavity The cancellous ends of the bones were quite fragile with fewer and smaller trabeculae With a long period of nonfunction in an adult animal the cortical bone of the shaft lost its compact struc

Related to this is the fact that engineers when calculating the trajectories of a structure assume that it is a solid homogeneous body, which an intact bone is not. Culmann's original diagram of the trajectories of a Fairbairn crane, which he compared with the trabecular orientation in the proximal end of the femur, was based on this assumption. Although this fundamental difference was recognized by Wolff (1870) he stated that the assumption of a solid body for outwardly lying tension and pressure lines leads to the actual conditions existing in a hollow section of a bone.

Triepel emphasized that the trabeculae did not always exhibit the orthogonality or 90° crossing as required by the Trajectorial Theory. Furthermore the individual trabeculae are often irregular and do not have the straight or regular curve as the theory demands. In trajectorial diagrams for engineering structures the pressure stresses are crossed by the tension stresses and, in the Trajectorial Theory of bone architecture, it is assumed that secondary trabecular systems crossing recognized pressure systems are tension resisting. However Triepel maintains that these secondary systems are not for simple tension resistance but are subjected to complicated bending actions. In addition he pointed out that bones may have similar external form and internal architecture although subjected to quite different strains and stresses in the living body. Although Triepel denies the trajectorial nature of spongy bone he admits that mechanical factors do influence and modify the trabeculae.

Triepel believed that the architecture of a bone is primarily dependent upon a geometrical relation between structure and form and only secondarily upon function. Thus bones of similar form should have similar structure regardless of function or the way they are stressed in the body. Vertebrae were cited as an example of this phenomenon. According to Triepel's interpretation the trabeculae in long bones consist of a series of domes and calyces fitting inside one another. The convexity of the domes is directed toward the epiphysis and the apex of the calyces away from it. The form and development of the domes varies within a single bone as well as from one bone to another. Other bony plates radially and horizontally oriented complete the arrangement.

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ture and became more porous. However, the general shape and contour of the bone as a whole was not markedly affected.

After long periods of nonuse during growth the bone was below normal size in length and thickness. The decrease in thickness was more pronounced in the shaft than in the epiphysis, resulting in a bone showing sudden enlargement in the epiphyseal regions. The decrease in the thickness of the cortex relative to the diameter of the shaft was small compared to that occurring in atrophy of bones in adult dogs. In the nonused bones the diameter of the medullary cavity, relative to thickness of the same bone, was always larger although the cavity itself was frequently smaller than the corresponding one in bones from the functional foreleg. Also the tibia, which was nonfunctional during the growing stage, had a circular cross section shape in contrast to the triangular shape of a normal functioning tibia.

Microscopic examination of sections of atrophied bone showed various stages of decrease in the number and size of trabeculae, thinning of the compacta, and accumulation of fat in the marrow. Ultimately there was an increase in the porosity of the compacta.

Wermel (1935a) completely or partially removed, with or without nerve section, one of the forearm or leg bones in young rabbits and rats and then studied the effects upon the size, form and surface markings of the remaining bone. Sectioning the nerve alone, so as to produce paralysis, resulted in disuse atrophy of the remaining bone but complete or partial removal without nerve section was accompanied by thickening of the remaining bone. This thickening was primarily the result of the increased stress upon the bone. However, thickening could also occur with nerve section and consequent loss of function. The thickening occurring in the latter case Wermel attributed to two factors: (1) the absence of normally existing pressure of the neighboring bone, and (2) an antagonism between the length and thickness of the bone.

In another paper (1935b) Wermel tried by measurements and mathematical formulae to relate changes in the thickness and cross section shape of bones to their resistance to compressive and bending stresses. However, he did not consider the density

of the bones and never experimentally determined the actual resistance of the bone to compressive and bending stresses Other experiments on the changes produced in bones by mechanical conditions of normal and pathological function, as well as the various interpretations advanced to account for them, are more fully discussed by Murray (1936)

The effect of immobilization on bones of the hind limbs of kittens and rats was recently studied by Gillespie (1954) Twenty eight kittens divided into three groups were used In the first group consisting of ten kittens the anterior roots of the lumbar and first sacral nerves were sectioned In a second group of ten kittens the posterior roots of the same nerves were divided In the third group of eight kittens lumbar sympathectomy was combined with section of the anterior roots of the same side and removal of all the lumbar ganglia from the renal vein to the pelvic brim

All the kittens in groups I and III exhibited significant reduction in the weight of the long bones and the thickness of the femoral cortex in the paralyzed limb However, no significant differences were found between the bones from the paralyzed and normal limbs in kittens of the second group

One hind limb in 16 young white male rates weighing approximately 120 gms was paralyzed by avulsion The animals were killed one week after operation and various physical properties of the femurs and tibias were analyzed In all animals the weight of the dry femur and tibia from the paralyzed leg was significantly less than that of normal bones

These experiments clearly show that the absence of functional mechanical stimuli cause definite changes in the gross anatomy of the bone However the question as to which is more important tension or compression in the development of normal bone structure is not answered

The functional significance of the architecture of compact bone has also been studied rather extensively Benninghoff (1925) developed the split line technique which consists of making a series of splits or punctures in the partially decalcified surface of a bone and rubbing India ink or water color into the splits The more superficial layers of bone are then further re

moved by decalcification so that split line patterns of the deeper layers can be obtained. Benninghoff used awls of various sizes for making the splits. Macerated and nonmacerated specimens of long and flat bones were studied and histological sections of the same bones examined under polarized light.

The split line patterns were reported to be constant for the long and flat bones. In the former the split lines coincided quite well with the long axis of the bone but quite complicated patterns were obtained in the flat bones.

The split line patterns were interpreted by Benninghoff as indicating *the orientation of the Haversian systems or osteones* which he believed lie in the long axis of tension and compression resistance. Therefore, the split lines indicate the direction of greatest tensile and compressive strength of a bone. Benninghoff did not, however, present any experimental evidence in support of his views.

Benninghoff accepted von Meyer's contention that the compacta corresponds to a closely crowded spongiosa and that the two types of bone simply represent different parts of the same stress system. Consequently, the orientation of the osteones is also trajectorial in nature. This idea was denied by Henckel cited by Murray (*loc. cit.*) who pointed out many instances, e.g., the proximal end of the femur, in which the split line pattern cannot be considered as a condensed continuation of the trabeculae of the spongiosa.

Another assumption of Benninghoff's was that the interstitial lamellae between osteones resist equally well forces impinging on them from any direction. He also believed that by their disordered arrangement the interstitial lamellae absorbed stresses in such a way as to make a bone statically and functionally homogeneous.

As previously mentioned (Chapters 8 and 9) split line studies have been recently made of the skull by Tappen and of the mandible by Dowgjallo and particularly Seipel. A modified Benninghoff method was used by Seipel (*loc. cit.*) on mandibles of individuals representing various states of health and disease. Although the split lines were considered as stress trajectories Seipel was well aware of the inadequacies of the technique and in con

trast to Benninghoff, illustrated some crossing of the split lines (Figure 32) He also emphasized that in bone adaptation the biological side of bone reaction must be considered as well as mechanical stimulation Thus he pointed out that changes in the form, function, elementary composition, nutritional, and vascular conditions of bone produce changes in its interior architecture and trajectorial qualities These are factors that do not seem to have been realized by others employing the Benninghoff method

Seipel reported changes in the trajectories of human mandibles as the result of variations in function arising from loss eruption or other changes in the dentition or from orthodontic appliances The orientation of the mandibular trajectories was considered to be the result of muscle action However the exact nature and magnitude of the forces presumably involved were not determined Seipel emphasized that from an anatomical viewpoint the tensile and compressive elements of a bone are not clearly differentiated because bone is contructed for multiple requirements

A simple mechanical interpretation of the architecture of the mandible was attempted by Seipel, who compared it with a straight rod or long bone (Figure 35), in which the upper trabecular systems represent the tensile and the lower ones the compressive trajectories The rod was then subjected to various bendings which displaced the tensile trajectories towards the outside and the compressive trajectories towards the inside In the adult mandible the tensile trajectories have an upper, anterior position and the compressive trajectories a lower, posterior location

Seipel also compared the alveolar arcade 4 in Figure 36 to an inverted bridge supporting the teeth and carried by the sloping basal tracts The inferior basal trajectories 1 were considered as compressive the oblique mandibular 2 and temporal 3 trajectories as tensile in function

Benninghoff's idea of the functional significance of the split line patterns was accepted by Kuntscher (1935a) who thought experimental confirmation of them was provided by the deformation patterns he obtained by static vertical loading of a colophonium coated femur However an opposite conclusion was

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tained in the colophonium coated bone revealed the true functional trajectories while the split lines represented growth trajectories arising from tensile strain created in the periosteum by growth of the bone at the epiphyses. Deviations in the course of osteones of bent bones are, according to Pauwels responses to changes in growth structure, as a result of bending rather than architectural adjustments to new strains in the bone.

A difficulty in accepting the mechanical sigfinificance of split line patterns is that the modern users of the technique apparently do not clearly understand the mechanical meaning of trajectories and stress. They write and speak of trajectories and stress as if they are visible entities which is not the case. A stress trajectory is the curve along which the principal stress at any point would fall and in order to draw such a curve the stress at the various points must be computed. Stress is the intermolecular resistance within an object to the deforming action of an outside force. Consequently stress cannot be seen but its magnitude can be computed provided the magnitude of the load and the cross section area of the object to which it is applied are known. If the

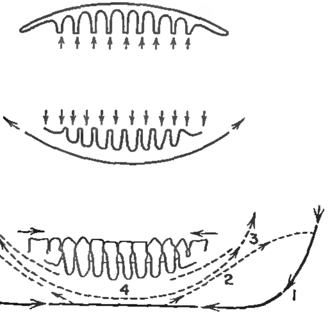

Figure 36 Alveolar arcade system of the mandible (From Seipel *Acta odont Scandinav* 8 1948)

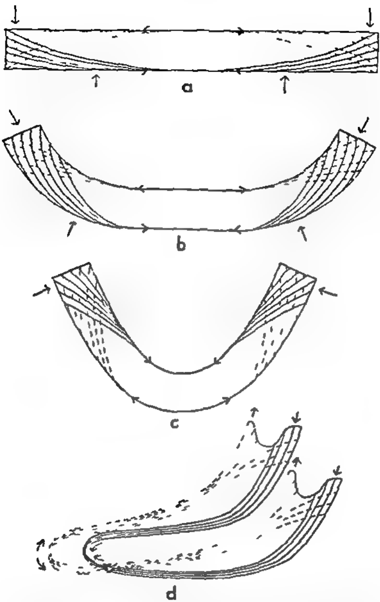

Figure 35. Simple trajectorial interpretation of mandibular architecture (From Seipel *Acta odont scandinav* 8 1948)

reached by Pauwels (1950) who compared the tensile and compressive trajectories obtained by similar loading of a colophonium coated femur with the split line pattern later produced on the same bone (Figure 37) Pauwels believed that the results ob

stress and strain in bones can be completely accepted Thus, are changes in the stresses and strains to which a bone is subjected accompanied by corresponding changes in the split line patterns? Do split line patterns change with growth of a bone? Are they present in a bone before birth or before many functional demands are placed upon a bone? These are some of the questions that should be answered by experimental evidence

The effect of immobilization, with consequent absence of mechanical stresses, on the microscopic structure of bone has been investigated by Engstrom and Amprino (1950) in the humerus and radius of dogs In a three year old dog the left front leg was firmly fixed to the body in a narrow skin pouch which prevented active movement Six months later the animal was sacrificed and x ray absorption studies made of bones from the inactive and normal limbs Similar studies were made in a second dog in which the left front leg was completely paralyzed by cutting the brachial plexus It was found that the ultrastructure of the active and inactive bones was the same although the latter bones weighed less than those from the normal limbs

Vigliani cited by Amprino (1951), made similar experiments on dogs, performing the operations at the beginning of the growth period so that bones from the operated and the control limbs would be entirely comparable He found that the histological structure of the bones from the immobolized limbs was identical with that of bones from the functional limbs Although there was less osseous tissue in bones from paralyzed limbs the osteones were fully developed and had been renewed in a regular normal manner Structural changes similar to those occurring in bones of normal human individuals have also been found in bones from limbs that have been paralyzed for about ten years

According to Amprino (*loc cit*) in a single section of compacta from any long bone which is obviously acted upon by equivalent mechanical forces the osteones vary greatly from one another The form length and distribution of the osteones is also modified throughout life Amprino believes that there are two possible explanations of bone architecture (1) that a bone achieves structural perfection, from a functional viewpoint only during a certain period of life or (2) that continual substitution

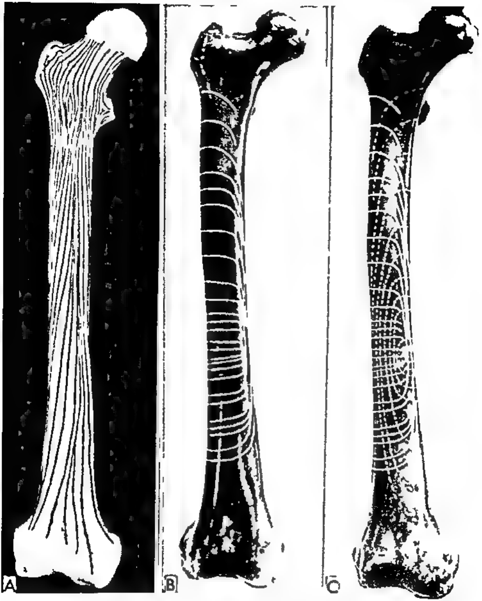

Figure 37 The split line pattern (a) compared with the pressure (b) and tensile (c—dotted lines) trajectories in a colophonium coated femur (from Pauwels *Anat Nachrichten 1* 1950)

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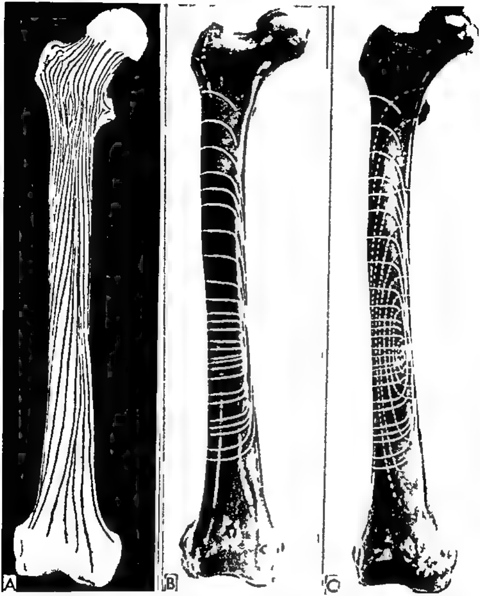

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a crane in frontal section and present an entirely different picture in a section slightly away from the midline, and (4) the basic nature of stress trajectories does not agree with the orientation of the trabeculae

There is considerable disagreement as to the nature of the stresses presumably responsible for bone architecture and growth, von Meyer, Roux and Wolff claiming it is tension, while Janssen and Carey say it is pressure In the author's opinion function, which may be in the nature of tensile or compressive stresses is the real stimulus for bone formation and growth The role of muscle action in the process is not clear The same is true of the functional significance of the patterns obtained by the split line technique Most of the ideas on the part played by stress and strain in bone formation and architecture are still largely theory and need support from experimental evidence On the other hand there is considerable evidence available to indicate that mechanical stimulation is not the only or necessarily the most important, factor responsible for the orientation and replacement of osteones

of different structures does not represent perfection from the viewpoint of mechanical resistance. In his opinion static and dynamic mechanical stimulations of a bone do not have a direct influence on the arrangement and structure of the osteones.

The influence of blood supply upon the architecture of bone was another factor ignored by Benninghoff. Ham (1952) points out that a bone cell, in order to survive, cannot be more than approximately 0.1 mm. distant from a capillary. Consequently, the size of an osteone, like the thickness of a trabecula of spongy bone, is limited by the necessity of having the bone cells near the blood supply rather than by mechanical stimulation. Also the general orientation of the osteones within a bone may be related to the course and arrangement of the capillaries. Amprino and Bairati (1936) also believed that Haversian systems should be considered in relation to growth processes and vascularization, i.e., nutrition, of bony tissues.

Summary

The preceding discussion shows that since the middle of the nineteenth century there has been considerable interest in the mechanical significance of bone architecture. At first most of the attention was centered on the spongy bone but, since the beginning of the present century, the compact bone has been investigated extensively.

The architecture of spongy bone has been most frequently interpreted in terms of The Trajectorial Theory, according to which the trabeculae represent trajectories adapted to resist the stresses and strains to which the bone is subjected. The Trajectorial Theory of bone architecture was most strongly supported by von Meyer, Culmann, Roux, and Wolff.

The Trajectorial Theory has been criticized on several grounds, the more important of which are: (1) Culmann's original trajectorial diagram of a Fairbairn crane, which was the basis for the theory, was drawn for a solid model composed of homogeneous material which is quite different from a hollow bone consisting of heterogeneous materials; (2) a femur is not loaded or subjected to stresses similar to those in a Fairbairn crane; (3) the femoral trabeculae only resemble tension trajectories of

placed into each of the holes and the two hooks connected with a rubber band under tensile strain. By varying the position of the drill holes in the flap and the temporal bones changes were produced in the distribution of the total force applied during an experiment. After the operation the skin and superficial fascia were closed.

At varying intervals following the operation the animals were sacrificed, the appliances removed and the entire flat surface of the calvaria excised and serially sectioned from behind forward. A combined total of more than 12 000 sections were stained with hematoxylin and eosin and examined microscopically.

By the described technique compression could be exerted upon the margin of choice. One free margin of the bone flap was compressed anteriorly but proceeding posteriorly, the degree of compression decreased and finally there was no contact at all. Thus material with contact and compression, with contact and excessive pressure, and with no contact could be observed. It was found experimentally impossible to ensure contact without some compression.

The "contact compression factor" is visualized as consisting of two parts: (1) contact of surfaces under compression and (2) a compression force endeavoring to bring into actual contact fracture surfaces which are only opposed. The compression may consist of an internal or an external force exerted on the fracture surfaces. An internal force is a physiological one created by muscles, while an external force is the result of nonphysiological factors such as gravity, weight bearing, and surgical techniques which may follow an anatomical pattern.

Of the 18 animals used in the study a contact compression factor was applied to 13 and omitted in five animals. Eleven animals with the contact compression factor were sacrificed after the fifteenth postoperative day. Of these 11 animals ten showed a favorable response to the force applied. None of the animals not subjected to the contact compression factor showed any significant osteogenetic activity.

The authors were aware of the many conditions affecting osteogenesis and, without altering or analyzing them, attempted to keep the physiology of the experimental fields of the control

11

The Effect of Stress on Bone Healing and Growth

THE ROLE of stress and strain in osteogenesis is of practical importance in the treatment of fractures and recently considerable interest has been shown in the influence of mechanical factors in fracture healing Most of the evidence of the importance of these factors is clinical in nature although a few experimental studies have been made

Compressive Stress in Bone Healing

Eggers, Shindler and Pomerat (1949) studied the role of contact compression in experimentally produced fractures of the skull of 18 mature white rats A longitudinal incision of the skin and superficial fascia was made over the middle of the skull and a three sided flap was cut in the left parietal bone One cut was made parallel to the sagittal sinus and a second parallel with it at a distance of 2 to 3 mm lateral to the first incision The two cuts were then connected by a third The resultant bone flap with its periosteum and blood supply intact remained attached behind providing a point of fixed anchorage

Contact compression was produced by passing cotton thread through holes drilled in the anteromedial and anterolateral part of the bone flap and the immediately adjacent area of the right and left temporal bones The thread was passed through the holes extradurally and extracranially and tightly tied If the thread were put through the two lateral holes lateral contact compression was created if through the two medial holes medial contact compression With lateral compression the margins of the medial cut were separated by approximately 3μ at the anterior hole with medial compression the opposite condition prevailed In medial compression a small stainless steel wire hook was

placed into each of the holes and the two hooks connected with a rubber band under tensile strain By varying the position of the drill holes in the flap and the temporal bones changes were produced in the distribution of the total force applied during an experiment After the operation the skin and superficial fascia were closed

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THE ROLE of stress and strain in osteogenesis is of practical importance in the treatment of fractures and recently considerable interest has been shown in the influence of mechanical factors in fracture healing. Most of the evidence of the importance of these factors is clinical in nature although a few experimental studies have been made.

Compressive Stress in Bone Healing

Eggers, Shindler, and Pomerat (1949) studied the role of contact compression in experimentally produced fractures of the skull of 18 mature white rats. A longitudinal incision of the skin and superficial fascia was made over the middle of the skull and a three sided flap was cut in the left parietal bone. One cut was made parallel to the sagittal sinus and a second parallel with it at a distance of 2 to 3 mm lateral to the first incision. The two cuts were then connected by a third. The resultant bone flap with its periosteum and blood supply intact, remained attached behind providing a point of fixed anchorage.

Contact compression was produced by passing cotton thread through holes drilled in the anteromedial and anterolateral part of the bone flap and the immediately adjacent area of the right and left temporal bones. The thread was passed through the holes extradurally and extracranially and tightly tied. If the thread were put through the two lateral holes lateral contact compression was created, if through the two medial holes medial contact compression. With lateral compression the margins of the medial cut were separated by approximately 3μ at the anterior hole, with medial compression the opposite condition prevailed. In medial compression a small stainless steel wire hook was

though its value was not determined It is very difficult to see what influence muscles would have in the conditions of the experiment because the magnitude of the force exerted by the involved muscles is unknown In addition it would vary with the state of activity of the muscle

The contact compression factor in bone surgery was emphasized by Eggers Ainsworth Shindler, and Pomerat (1951), who pointed out that compression arthrodesis, alterations of the shelf in hip operations, excess bony proliferation on articular margins and bone grafting by muscular pressure so as to retain the graft against the recipient bony surface are adaptations of the contact-compression factor Fractures should be treated so as to utilize this principle and convert shearing forces into pressure forces for good clinical results They believe that in children operative treatment of fractures is rarely indicated because closed reduction utilizing the contact compression factor, is associated with growth and will result in good satisfactory recovery In the treatment of scoliosis by vertebral fusion, they showed that fusion will be greatest on the concave side where pressure is the stronger In all such techniques optimum pressure is striven for because excessive pressure causes necrosis of bone rather than osteogenesis Presumably the magnitude of the optimum pressure varies according to the fracture involved

Compression arthrodesis of the knee was discussed by Charnley and Baker (1952) on the basis of operations performed on 67 knees in 63 patients Four cases were bilateral Compression was applied for approximately four weeks after operation and then walking permitted in the plaster cylinder Osseous union was achieved in 63 of the 67 cases The histological nature of the union was verified by a biopsy specimen The end result of all cases was followed by clinical and radiological tests

It was frequently found when the compression produced only moderate degrees of rigidity, that a solid knee was often obtained in four weeks The authors interpret this as indicating that the essential mechanical feaure is not absolute fixation but an absence of shearing movement They also point out that cancellous bone united more readily than compact bone for the following reasons (1) osteoblastic action is confined to bone

and of the affected part virtually identical This was done by conducting both in the same animal within a limited transverse physical space of about 2,500μ Thus in the experiments the control margin can be seen simultaneously with the experimental margin of the bone studied Under these conditions they assume because of such close proximity, that no great physiological varia tion was present

During the experiments it was found that those animals sacri ficed before the sixteenth postoperative day show little or no osteogenetic activity In one animal sacrified on the ninty fourth postoperative day, nonunion was found The significance of this is impressive when contrasted with the 17 day sections in which *osteogenesis and union were present on the side of the flap to* which the contact compression factor was applied

The consistency of the results obtained convinced the authors that the only reasonable variable was the contact compression factor They found that union could be secured on either the medial or lateral aspect of the bone flap by choosing the side to unite and then applying the contact compression factor If they wished no union the contact compression factor was not applied They also found that those margins of the bone flap to which ex cessive pressure and contact were applied became necrosed with a delay in osteogenesis However, excellent osteogenetic activity was observed in those margins under more ideal pressure and contact In the approximated margins with pressure applied marked tendencies to bridge the intervening fracture gap were evident Thus they found that the contact compression factor with too much force caused necrosis but there was no osteo genetic response when the factor was absent The authors be lieved that the optimal pressure for osteogenesis was somewhere within the physiological limits of the force exerted by the muscu *lature of the individual concerned*

One of the chief criticisms of these experiments is that the actual magnitude of the force exerted on the bones during the experiment was not determined It was shown that while too much pressure caused necrosis no osteogenesis occurred at all if too little pressure were applied Thus there is an optimum magnitude of pressure necessary to produce osteogenesis, al

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surface, endosteal or periosteal, (2) the total endosteal surface is very great in cancellous bone whereas in compact bone it is limited to the linings of the Haversian canals and (3) the blood supply is proportionately greater in cancellous bone. Compression arthrodesis has been used by others and is now a fairly common surgical procedure

Charnley and Baker did not indicate the magnitude of the pressure used in the arthrodesis but according to Reynolds and Key (1954) it amounted to a force of approximately 25 lbs/in² The latter authors also reported their results from an experimental study on fracture healing in which fixation was obtained by the use of standard plates contact splints and medullary nails Experiments were made with 40 adult dogs divided into two groups In the first group of 19 dogs one femur was divided transversely in the middle third, the fracture reduced as accurately as possible and the fragments fixed with a standard steel plate with six screws The other femur was operated on in a similar manner and fixed with a stainless steel Eggers plate with six screws In the second group of 21 dogs similar osteotomies of the femur were performed The left femur was fixed with an Eggers plate and the opposite femur with a medullary nail At intervals of seven to 16 weeks the animals were sacrificed and the femurs removed and examined by gross inspection, x ray and microscopic studies

In both groups somewhat better results were obtained with a contact splint This was most marked in comparison with medullary nails and with the standard plate They felt the experiment demonstrated that fixation and apposition of the fracture fragments are more important in obtaining union than the pressure of weight bearing Again the magnitude of the pressure exerted at the fracture site was not determined In the discussion of this paper Eggers pointed out that bone is only absorbed and not replaced in the absence of compression When compression is present however osteogenesis and absorption are going on simultaneously

The effect of pressure on the healing of bone grafts was experimentally studied by Ford Lottes and Key (1951) in the ilium of dogs Sections of the lower two ribs on one side were

removed subperiosteally and transplanted with the convex surface facing outward, into each ilium The rib grafts were about 5 cm long On the left side two holes were drilled obliquely in the ilium so that one end of the rib graft could be placed in each hole Considerable force was required to force the graft in place and it was held in position by its own elasticity When in place the graft exerted considerable pressure on adjacent portions of the ilium

On the right side a hole was drilled obliquely in the posterior part of the ilium while a groove was cut in the anterior part so that when one end of the graft was placed in the hole the other end rested in the groove without undue pressure The graft was laid loosely in position with the convex surface outward, and held in place by suturing soft tissues over it

The eight dogs used in the study were sacrificed at one, two, three four, five six eight and 19 weeks postoperatively Each pelvis was incised and the ilium, with the graft in place, was examined roentgenographically On the left side the section of the ilium between the ends of the graft was cut in order to determine whether the elasticity of the graft was still sufficient to separate the fragments of the ilium In no case was this true Therefore the pressure which the ends of the graft exerted on the adjacent bone must have decreased considerably within a few days after the placing of the grafts in position

Gross examination showed that on the left or pressure side the ends of the graft were embedded in the holes in the ilium and appeared to be firmly fixed even after one or two weeks On the right or nonpressure side the ends of the grafts were united only by fibrous tissue at one or two weeks postoperatively After three weeks bony union between the graft and the ilium was present on each side and became progressively firmer in experiments of longer duration there being no perceptible difference between the pressure and the nonpressure grafts

Roentgenographic examination showed that new bone first appeared on the outer surface of the ilium adjacent to the graft It was visible in all specimens taken more than three weeks postoperatively Clinical union by bone appeared to be present on each side after four weeks and in all older specimens No differ

ence was noted between the pressure and the nonpressure sides

Microscopic examination of sections through the junction of the graft and the host on the pressure side showed evidence of union of the two by new bone In every instance in which the origin of new bone could be identified it arose from the host bone No detectable growth occurred from the graft which soon died A similar change occurred in a rather wide zone around the drill hole in the cortex of the ilium One week postoperatively practically all the marrow and the rib grafts were dead and absorption of the graft by osteoclasts had begun By two weeks the absorption was well advanced Delicate trabeculae of new bone were already present in specimens removed after one week of implantation of the graft in the ilium *The trabeculae arose from the* cancellous bone of the ilium

At the pressure areas where the graft cortex was opposed to and pressed against the margin of the freshly cut hole in the iliac cortex two apparently dead bone surfaces were in contact On the nonpressure side where one end of the rib was inserted loosely into a relatively large drill hole while the other was placed in a fairly wide groove in the ilium the space between the graft and the host bone was usually wider than on the pressure side *However this seemingly caused no appreciable delay in the union* of the two

One week postoperatively delicate new bone trabeculae apparently arising in the cancellous bone of the ilium were present The amount of new bone appeared to be greater than on the pressure side because the spaces to be bridged were wider In addition to the new bone the trabeculae of the five or six weeks postoperative specimens exhibited small areas of hyaline cartilage in the new bone binding the graft to the ilium at the front The cartilage resembled the ordinary callus formed in the process of fracture union No cartilage callus was seen in any of the specimens from the pressure side but in other respects the differences between the two sides were only minor

The authors believe the beneficial effects of pressure on bone union are due more to complete immobilization of the bone affected by the pressure than to the pressure itself If immobilization is complete callus is more abundant and more likely to con

tain cartilage has occurred in some of the experiments on the pressure side Pressure exerts a more favorable influence on cancellous than on compact bone because of interdigitation of the trabeculae in the former However, pressure made little difference in the manner or rate of union

In contrast to the preceding studies are those of Friedenberg and French (1952) in which the effects of known compression forces were studied Compression was applied by means of calibrated springs placed across a fracture created between the proximal and middle portion of the ulnas of dogs The distal third of the bone was resected as it was necessary to remove the interosseous membrane of the middle fragment so the pressure could be applied effectively A threaded stainless steel wire was inserted through the olecranon and down the medulla of the ulna The wire crossed the fracture line and passed completely through the middle third of the ulna and through the calibrated spring The wire was fixed with a nut at the olecranon and a second nut was used to tighten the spring thus forcing the middle ulnar fragment against the proximal one The fracture ends were first approximated without pressure and the spring was measured with calipers Further pressure of the spring was accomplished as desired and the spring length measured again The difference between the two measurements represented the spring compression acting to force the loosened middle fragment of the ulna against the proximal one The springs delivered a force of 6 lbs for each 0 031 inch of shortening

Four to five weeks postoperatively roentgenograms were taken and the fracture site exposed in order to determine the callus development and the presence or absence of mobility The spring was again measured to note what loss of pressure had occurred The fracture area was then excised and sections made for microscopic examination

The initial pressure created across the fracture *did not remain* constant throughout the experiment Pressure was lost by the tendency of the spring to return to its original length and by bone absorption most of which occurred adjacent to the spring In addition, an unknown loss of pressure occurred because of

ingrowth of fibrous tissue into the spring and a fixation of bone fragments by scar tissue

Twenty seven experiments were completed on adult dogs weighing *from 17 to 25 kgs Usually the dogs began to bear* some weight on the operated extremity by the fourth week However little or no weight bearing stress occurred across the fracture because of the resection of the distal third of the ulna and the interosseous membrane attached to the middle fragment

For comparative purposes the experiments were divided into four groups according to the magnitude of compression originally placed across the fracture Pressures between 30 and 36 lbs constituted the high group, those between 12 and 18 lbs the middle group those between 5 and [illegible] lbs the low group, and those without any pressure but simple intermedullary wire across the fractures were the fourth or control group

In the five specimens of the high compression group there was only one case of union This occurred in an animal that sustained a 15 lbs loss of pressure, and showed marked bone absorption adjacent to the spring The average pressure lost in this group was *10 lbs* In most instances there was scant soft tissue growth between the surfaces and reduced perosteal proliferation of tissue adjacent to the fracture

In the intermediate group four cases of union were found in six experiments Three of the four animals observed five weeks postoperatively had union and half of those observed at four weeks postoperatively showed union An average measured pressure of six pounds was lost during an experiment

Two of ten experiments in the low pressure group pointed to bone union No measurable pressure loss was ascertained and the position of the fragments was well maintained In contrast to higher pressure groups soft tissue proliferation about the fracture was abundant

In the six control experiments only one animal developed bone union five weeks postoperatively Although alignment of the bone fragments was adequate it was not as satisfactory as that *in the other* groups because of slight transposition movements of the fragments along the medullary wire

Microscopic examination of the fractures of the higher press

ure group showed little cellular reaction in the fracture gap in most instances Occasionally degenerative changes, with osteoclasis at the fracture, were noted for the cortical bone New periosteal and endosteal bone were found at a distance from the fracture with the periosteal collar more prominent

In the intermediate pressure group new medullary bone was as abundant as new bone on the exterior of the cortex New endosteal bone was condensed along the inner surface of the cortex form ing an inner core of new bone oriented along the medullary canal toward the fracture Profuse fiber and cartilaginous formation was observed in localized areas in and around the fracture gap

Microscopically the fractures in the low pressure group showed conditions not qualitatively different from those in the intermediate group However, the state of development of the new bone was generally not as far advanced as in the middle pressure group In the control fractures active periosteal and endosteal bone formation was observed with islands of hyaline and fibril cartilage centered around the fracture gap

The healing process closely followed the observations of Urist and Johnson (1943) with some variation in the different pressure groups In the low and middle pressure groups very substantial endosteal new bone formation occurred The fractured cortical surfaces seemed to contribute cellular growth only late in the healing process Aside from a retardation of the cellular activity adjacent to the fracture area and late necrosis of the cortical ends in the high pressure group no specific alteration in the process of fracture healing which could be attributed to pressure was observed In the middle and low pressure groups more rapid healing occurred than in controls possibly as a result of more rigid fixation and close bone contact

Friedenberg and French point out that in any fractured surface regardless of whether it is in a tubular or flat bone the pressure across the surface is primarily borne by cortical bone Therefore any specific reaction to compression should be sought in the cortical bone No attempt was made to determine what effect mechanical forces applied to the fracture had on such factors as the local pH, phosphatase calcium phosphorus, or other biochemical responses incident to a fracture

In the ordinarily treated fracture of a long bone Friedenberg and French emphasize that a compression force is exerted by the tone of the muscle groups surrounding the fracture In their study union was accelerated by increasing the pressure to a range between 12 and 18 lbs but they did not attempt to determine the magnitude of the physiological pressure presumably exerted by muscle tone across the fracture However, they believe, even in the low pressure group, that the physiological pressure would be less than that produced by the rigid bone impaction they observed at the time of surgery Raising the pressure force above that which the muscles were capable of applying seemed to accelerate healing until a point was reached where cellular response lagged They believe a curve of bone union plotted against pressure would show the summit of the curve well above physiological pressures

Compressive Stress in Bone Growth

The influence of mechanical factors on bone growth in the tibia of newborn Holstein calves has been studied by Strobino French and Colonna (1952) Two different devices were used with the object of determining the force exerted by the growth of epiphyseal bone and more particularly the rate of growth of bone under increasing tensions However as will be apparent later they did not achieve their objective

The first apparatus used consisted of two heavy steel coil springs each encased in a cylinder closed at the ends with stainless steel plugs screwed into position A hole in the top plug permitted the free passage of a hooked rod, while the lower plug had two small holes to admit restraining wires The end of the rod inside the cylinder was held in place by a washer and a nut The spring was greased with vaseline for lubrication and water proofing purposes A heavy pin was then driven through the epiphysis and another through the diaphysis about a third of the distance from the proximal end of the tibia The rods in one end of the spring cylinder were then hooked over the epiphyseal pin while the opposite end of each cylinder was connected with the diaphysial pin by heavy wires threaded double through holes in the pins

The whole apparatus was placed so that the spring on each side was as close as possible to the tibia Relatively high voltage x rays were taken at regular intervals to visualize the compression of the springs within the cylinders The springs were calibrated so that one inch of deflection equalled a total force of 60 pounds The limit of the apparatus was 120 lbs, i e 60 lbs for each spring

In one bull calf six weeks of age the spring tension at the beginning of the experiment was zero pounds During the course of the experiment the springs gradually became compressed as the force across the epiphyseal line increased At the end of the eighth postoperative week a force of 75 lbs, equivalent to approximately 10 lbs/in² of epiphysis was present and the retaining wire on one side ruptured thus releasing the tension on that side The animal was followed for another month during which "the tension on the unbroken side in which the apparatus was intact increased from 37 lbs to 52 lbs However, the growth rate was practically the same on both sides of the tibia When the growth of the bone (inches) was plotted against time (days) a straight line was obtained Such a curve was called a growth curve under increasing tension

In a second calf six weeks of age essentially the same apparatus was used except the initial force was set at 65 lbs by compressing the spring to the required degree When the 120 lb level had been obtained after a period of about three months the apparatus locked, although growth continued until the two stainless steel anchoring wires on each side were stretched and broken almost simultaneously The breaking strength of these four wires had been previously determined to be 70 lbs per strand Thus the growth of the epiphyseal plate continued up to 280 lbs of wire tension Although the wires ruptured at this point the epiphyseal growth continued unabated The experiment lasted for a period of five months taking two additional months to stretch and break the retaining wires after the coil springs had become firmly compressed and locked

Since this experiment showed that the force exerted by growing bone was several hundred pounds, it was found that the spring tension devices were no longer practical Therefore the second apparatus was based on the stress strain relationships of

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lease of tension At the end of six months both tibias were exactly the same length

In an experiment on a calf eight weeks of age the stainless steel wires were used in place of the springs Four strands of wire, each 0 028 inches in diameter and with a combined strength of 560 lbs, were used on each side During the experiment the slack in the wires was taken up by the growth of the bone and the epiphyseal pin was gradually pulled down into the diaphysis Although the wires did not break their stretching indicated a force of approximately 400 lbs In spite of this the tibia grew at the same rate and amount as on the control side The growth curve was a straight line and there was no slowing of growth with this high force (400 lbs) which exceeded the weight of the animal The authors stated that the pin across the epiphysis was in contact with the bone a length of approximately four inches and had a cross section of 0 210 inch this resulted in a cross sectional contact area of 0 84 inch or 475 lbs to the square inch The pin in the mid shaft was in contact only with the relatively thin cortex on each side of approximately $^3/_8$ inch and had a cross section of 0 190 inch They determined that this yielded a contact pressure of approximately 2 500 pounds per square Although a moderate overgrowth of bone on the contact sides was seen there was no roentgenographic evidence of pressure necrosis The authors very probably meant 2500 lbs/in^2 as 2500 pounds per square has no significance

The authors concluded that the growth rate of bone subjected to a pressure of 400 lbs or 60 lbs/in of cross section, was exactly the same as on the contralateral side which was growing under normal physiological tensions They also emphasized that the force acting on the bone was well beyond what would be expected of physiological forces such as muscle pull or the weight of the animal at the time the reading was made The rate of growth of the epiphysis was in no way altered when the epiphyseal plate was released from a prolonged force up to 280 lbs which had gradually been increasing over a period of several months

The results and conclusions of Strobino, French and Colonna do not conform with the title of their paper *The Effects*

stainless steel wires The force elongation relationship (modulus of elasticity under stretching) of the wires was determined experimentally up to the breaking point and served as a calibration curve for evaluating the unknown force from the measured elongation Molybdenum stainless steel wires with an initial diameter of 0 028 inches and a breaking strength of 70 lbs were used At the breaking point the wire had a diameter of 0 023 inches and a stretch of about 40% of its initial length If the wire is stretched beyond its yield point the additional force necessary to produce the same increment becomes progressively less until the breaking point is reached By using multiple parallel strands of wire a very large force can be measured, the stress strain curve for the wire holding true in direct proportion to the number of strands used A battery of four parallel strands of the wire had an ultimate tensile strength of 280 lbs The stress strain curve was the same for four strands of wire as for one By this method the force exerted against the wires being stretched could be measured with a relatively high degree of accuracy

The growth measuring device consisted of a slider type apparatus formed from a thin rod bent at a right angle The short arm of the rod was inserted into the epiphysis while the long arm extended downward with its lower end through a hole in a plate attached to the upper tibial shaft As growth occurs the rod slides up its protruding end giving an accurate measurement of the magnitude of the growth During an experiment the lower end of the longer arm was eventually drawn up beyond the plate so that the distance from the tip of the rod to the plate could be easily measured By this method a growth increment of as little as 0 04 inches could be measured The authors stated that the apparatus was placed on the contralateral side of the tibia of each animal

In the first animal the growth on the tension side was at the same rate as on the opposite side and gave a straight growth rate curve After the retaining wires were broken on both sides of the tibia the growth measuring device was inserted on the tension side During the subsequent month the growth rate was identical on both sides and was in no way altered after the re

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of Increasing Tensions on the Growth of Epiphyseal Bone or the stated purpose of their investigation In their experiments the springs and wires connecting the epiphyseal and diaphysial pins, not the bone were subjected to gradually increasing tensions by the growth of the bone The bone itself was under compression because the restraining effect of the springs and wires between the pins tended to resist their further separation as the result of bone growth

The experiments actually showed that epiphyseal bone growth was not stopped by compressive forces of 475 lbs/in^2 and 2500 lbs/in^2 developed under the epiphyseal and diaphysial pins, respectively Even in the experiment in which the spring of one side was broken the remaining one would exert a compressive not a tensile, force upon the bone of the same side The authors measured the effect of increasing compression not tension, upon the growth of epiphyseal bone

Tensile Stress in Bone Growth

The influence of tension upon the growth of endochondral bone of young healthy dogs was investigated by Gelbke (1950) Tension was exerted by fixing the elbow in the position of extreme extension by means of a wire passed through holes drilled in the olecranon epiphysis and the middle of the humeral shaft Thus the olecranon epiphysis was placed under tension while the distal humeral epiphysis was subjected to compression The opposite normal elbow served as the control

During an experiment the length of which varied from four to 15 weeks the tensile stress on the olecranon was slowly increased by the growth of the forearm and the action of muscles Examination of the control limb showed that the operative trauma and the reaction to the wire could be neglected At various postoperative intervals x rays were taken of the operated and control elbow in order to examine the growth of the endochondral bone The animals were later sacrificed and the olecranon region studied histologically

During the experiments no visible deformations or increase in growth of the olecranon were produced However roentgenographic examination revealed rarefactions and indistinct de

maturation of the epiphysis accompanied by cyst-like rarefactions of the metaphysis. Bony infiltration started with narrowing of the epiphyseal cartilage which in some areas was so marked that the spongiosa of the diaphysis and metaphysis blended. Irregularities in the course and cell order were also seen. Intensified cartilage absorption, in favor of bone formation, was found in the operated animals as well as in those with almost completely grown long bones having partially ossified diaphyses.

Some information regarding the effect of pressure was also found in the distal humeral epiphysis. Thus x-ray and histological examination revealed local interruptions, narrowing, and bony infiltrations of the epiphysis on the operated side. The humerus of the operated side elongated several millimeters, starting from the epiphysis which was already subjected to increased physiological growth. This was attributed to stimulation by the wire through the shaft, as a dense, ring-like bone formation was noted around the canal for the wire. At this point transformation of the spongiosa, probably as the result of mechanical stress, had occurred. These results confirmed and supplemented Gelbke's previous investigations.

On the basis of his experiments, in which a strong, continually acting mechanical stimulus was used, Gelbke made the following conclusions regarding the role of tension and pressure in bone growth:

1. Strong compressive forces hinder endochondral bone growth, although strong tensile forces do not increase growth.

2. Both tension and pressure, in intensified continuous action, cause disappearance of the epiphysis after gradual narrowing.

3. Therefore, the growth of the particular skeletal region involved ceases under tension.

4. Continuous tensile and compressive forces of equal magnitude have the same effect on the processes of endochondral growth and ossification. The epiphyseal material which is sensitive to mechanical stimulus and is rapidly growing is replaced by spongiosa which is resistant to tensile and compressive forces. Thus, a bone, in order to retain its configuration and stability, sacrifices its growth organ and thereby its possibility of growth.

The results of Gelbke's studies are not in accord with ideas expressed by earlier investigators that tension or pressure, depending on the author's viewpoint stimulates or retards bone growth In Gelbke's experiments tension on the olecranon epiphysis and pressure on the distal humeral epiphysis both caused replacement of the epiphyseal material by spongiosa resistant to mechanical stress This process occurred simultaneously in the same animal

Summary

Both clinical and experimental evidence indicate that pressure or compressive stress can stimulate formation of new bone an important factor in fracture healing However the experiments reveal that this is only true up to a certain point and that too much pressure causes necrosis and bone destruction. Friedenberg and French (*loc cit*) showed that a pressure range of 12 to 18 lbs accelerated bone growth in dogs whereas only one case of union occurred when the pressure was 30 to 36 lbs Even in this specimen the pressure had dropped to about 15 lbs

The influence of pressure on bone growth varies with different species the tibia of Holstein calves withstanding compressive stresses of 475 lbs/in² and 2500 lbs/in² without cessation of epiphyseal growth No pressure necrosis was noted although it has been found in rat skulls and the long bones of dogs

Continuous tension over a relatively long period of time stops bone growth in the dog and causes the replacement of epiphyseal material by more resistant spongy bone Compressive stress has a similar effect if allowed to act continuously for a time Strong compressive stress hinders endochondral bone growth in the dog although tensile stress does not increase it

The experimental results described seem to indicate that compressive stresses may be a somewhat more important factor in stimulating bone growth than tensile stresses However the effects of compression vary in different animals and too much compression can cause necrosis The optimum amount of pressure for fracture healing has not been definitely established for man There is also evidence that immobilization may be as important as compressive stress in fracture healing

12

Stress and Strain During Embryonic Development

THE INFLUENCE of mechanical factors in osteogenesis and the development of bone form is of considerable importance. For example, what role do mechanical stresses and strains play in the development of ectopic bone or some of the skeletal abnormalities often found at birth? How is the postnatal form of bones modified by mechanical stresses? Since the clinical evidence is inconclusive, the answers to these and similar questions must be sought in the field of experimental embryology.

Effect of Stress in Embryonic Osteogenesis

The influence of tensile and compressive stress on osteogenesis in embryonic and newly hatched chicks has been experimentally investigated by Landauer (1927), Murray and Selby (1930), Studitsky (cited by Murray 1936) and Glucksman (1938, 1942). Landauer studied the skeletons of chondrodystrophic embryos. Murray and Selby made chorio-allantoic grafts of complete or fragmented limb buds from four and five day chick embryos as well as the cartilaginous femur of six day embryos. Similar material was used by Studitsky. Glucksman investigated osteogenesis by the hanging drop method in embryonic skeletal rudiments or endosteal cultures from late embryonic or newly hatched chicks.

In the grafting experiments the long bones were frequently bent, presumably by the resistance of the chorio-allantois to their growth in length. Similar bendings and even fractures, probably resulting from muscle pull, occurred in the long bones of the chondrodystrophic embryos. In all cases, regardless of the possible cause of the bending, there was hypertrophy of bone on the concave side so that the curvature was reduced. Even in nor

mally curved femurs of 15 day chick embryos there was more bone on the concave than on the convex side of the shaft so that the femoral curvature was less than in the cartilaginous stage

In the hypertrophic mass of bone on the concave side the trabeculae arise from the thin sheath of bone investing the cartilage and extend obliquely toward the fibrous layer of the periosteum Landauer attributes this trabecular arrangement to the extrinsic compressive forces that caused the bending in the bone If this is true, as pointed out by Murray (1936), mechanical principles dictate that the trabeculae should proceed along the concave side of the bone because in the region subjected to the greatest pressure, the lines of principle force run down the concave side Therefore, the radial orientation of the trabeculae indicates there is no direct relation between the direction of the force bending the element and that of the trabeculae

Studitsky attributes the radiating arrangement of the trabeculae to two series of tensile stresses which arise in the inner angle of flexure and are created in and below the periosteum by growth of the cartilage One of the series unites the wings of the cartilage which diverge with further growth, while the second series, arising as a result of the withdrawing of the first from the inner angle of the bend unites the fibers of the first series with the vertex of the angle of the flexure This idea does not, however, agree with the observed architecture of the bone in question

In other experiments Studitsky removed the perichondrium from the shaft of cartilaginous chick long bones wrapped them in periosteal fragments taken from human fetal long bones and then made *chorio allantoic grafts* Later examination showed that the human periosteum formed an osseous investment around the shaft of the cartilaginous chick bone just as does the normal chick periosteum If the chick cartilage were bent the human bone on the concave side had the same radiating trabecular arrangement as would be formed by chick bone in the same place This seems to indicate that the cartilage model does have some mechanical influence on the periosteum as Studitsky believes

Studitsky and Landauer apparently believe that the new bony architecture arises as a direct response to new mechanical conditions acting on the bone . However, Fell (1925) demonstrated

in longitudinal sections through the tibia of late embryos or newly hatched chicks that the trabeculae in the shaft normally have an oblique orientation passing from the inner bony sheath around the cartilage out to the periosteum. A similar arrangement of the trabeculae, as if radiating from a center, is also present in the long bones of adult chickens. Thus, the oblique or radiating orientation of the trabeculae on the concave side of bent bones is a consequence of ossification along normal lines and not a special adaptation to extrinsic mechanical forces acting on the bone.

Murray (1936) points out that even the hypertrophy of osseous tissue on the concave side of a bent bone is not necessarily the result of increased compressive stresses. In the early stages of ossification of a long bone the cartilage model has an hour-glass shape, while the periosteum is just attached to the bone at each end. Consequently, except near the ends of the bone, there is a subperiosteal space, widest in the region of the middle of the shaft, on each side between the periosteum and the cartilage. During development these subperiosteal spaces become filled with compact bone. The existence of the subperiosteal space depends on the shape of the cartilage model. Consequently, if the model is bent, the subperiosteal space on the convex side is almost entirely eliminated while that on the concave side is widened. When the latter space is filled in it represents the hypertrophied bone discussed by Landauer and Studitsky. The amount of bone present on each side of a developing long bone thus depends upon the width of the subperiosteal space on each side of the shaft.

The latter idea was experimentally verified by Murray and Selby (*loc. cit.*). They found that chorio-allantoic grafts of intact femurs from six-day chick embryos produced a large quantity of bone, while in similar grafts of halves of femurs from seven-day chicks just a thin shell of bone was formed. They accounted for the latter condition on the basis that in the half femurs only one epiphysis was present so that the specimen was wide at that end but tapered to a blunt point at the other. Consequently the periosteum was not lifted away from the underlying cartilage as

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the ribs and elongated slightly parallel with the ribs. After different periods of growth the cultures were fixed, decalcified, and studied histologically.

In well ossified cultures three days after implantation, the original ossified ring was slightly flattened at the site of contact with the ribs while the entire culture was elongated parallel with the ribs. A new structure consisting of bony trabeculae oriented perpendicular to the ribs was usually present. This resistance structure persisted in 12 cultures from the third to the eighth day after implantation between the ribs and prevented their further approximation.

The observations on living material were confirmed and amplified by histological examination of fixed cultures. The outermost bony ring corresponding to the original area of outgrowth retained the concentric cell and ground substance arrangement except where the main trabeculae met it. The culture as a whole showed little compression except for slight flattening adjacent to the ribs.

Slightly ossified cultures, the third day after implantation, were more elongated parallel with the ribs and the concentric rings were broken open in the same direction. Calcified trabeculae oriented parallel with the ribs formed a compression structure in the interior of the culture. This calcified implant usually prevented further approximation of the ribs. However, with less intense calcification approximation of the ribs might continue for a few more days. The presence of many osteoclasts filled with bone debris indicated that extensive resorption and reconstruction was also in progress.

In slightly more calcified specimens a structure intermediate between the resistance and compressive types appeared the third day after implantation. It consisted of small trabeculae at first perpendicular to the ribs but during the next few days they thickened and the pressure of the ribs shifted them to an acute angle to the direction of the ribs. The general outline of the culture was also flattened. A disturbance and break of the concentric arrangement of the ring formed by the original outgrowth occurred at the point where the trabeculae united with it.

To determine whether there were any other effects except

in an intact bone and the subperiosteal space was virtually eliminated Thus, only a small amount of bone was formed

This interpretation would explain the normal development of the thickest bone in the center of the shaft hypertrophy of bone on the concave side of a bent element and the deficiency of bone in the absence of one epiphysis No extrinsic mechanical forces are necessary The resulting bone form is a growth structure rather than an adaptive modification although the additional bone in the side subjected to maximum compressive stress has an important functional significance Similar changes occurring in an adult bone would probably be considered as modifications in adaptations to new stress conditions However, as Murray suggests many of the so called adaptive modifications may actually be growth phenomena instead of a direct response to functional mechanical requirements

The influence of pressure on the orientation of structure in *endosteal cultures from tibias of late embryonic and newly* hatched chicks was investigated by Glucksman (1938) Since this material normally does not exhibit oriented structure it is very suitable for studying orientation as produced by mechanical factors

Pressure was applied to the endosteal cultures by implanting them at various stages of ossification and calcification on the intercostal muscles between adjacent ribs explanted in vitro The ribs and intact intercostal muscles were taken from 21 different chicks ranging in age from the seventeenth day of incubation to the fifth day after hatching The preparations were cultivated by the hanging drop method

After a latent period of varying length the explants contracted so that after ten days or less the intercostal muscles had degenerated and the two ribs were in contact In coming together the ribs exerted considerable pressure on the endosteal implant which was sometimes bent or broken

The pressure stresses caused by the approximation of the ribs were investigated by cutting a square hole in the intercostal muscles on the third day in vitro and filling it with plasma The explants were then transferred to fresh medium within 24 hours during which the hole narrowed in a direction perpendicular to

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those due to pressure cultures were made of the lateral part of the ribs including the uncinate process Approximation of the ribs was prevented by fixing the uncinate process between them while the endosteal culture was placed in the intercostal space Although the muscles degenerated in the usual way the ribs re mained in their original position Thus pressure was the only factor present in previous experiments which was excluded from this one

Glucksman found no orientation whatever in the 20 untreated endosteal cultures but all 49 of the cultures subjected to pressure showed a compressive or resistance type of structure The former was oriented along lines of tension the latter along lines of pres sure In slightly ossified cultures the original bony ring fractured along tension lines which provided an outlet for the material within the ring This caused orientation along tension lines and the formation of the compression structure In well ossified cul tures no such outlet was possible because the ring broke, if at all, in the pressure lines Consequently material was pushed inside the ring by pressure from the ribs and condensed along the lines of pressure The presence of osteoclasts indicated an active remodeling of the bone structure was also occurring The degree of calcification when the cultures were first subjected to pressure determines whether a compression or a resistance struc ture develops

Glucksman (1938) also investigated the influence of tensile stress by subjecting the zone of outgrowth of endosteal cultures of chick bone to tension This was accomplished by removing the original bone and allowing the retracting clot to pull on the ring of outgrowth The result of the experiments revealed that the orientation of the osteogenic fibers depends upon the tension to which they are subjected In the presence of tension the fibers are parallel to each other but in the absence of tension their orientation is independent of one another

In a later paper Glucksman (1942) continued his studies of the effect of tension on osteogenesis in skeletal rudiments of chick embryos grown by the hanging drop and watch glass methods The first series of studies consisted of 54 experiments using fe murs and tibias of four day embryos and metatarsals and pha

langes of seven to 12 day embryos "Barrier" rudiments, placed in the direction of expansion of the explant being studied prevented the normal elongation of the latter and caused it to bend Thus the convex side of the bent explant was subjected to tensile and the concave side to compressive stress The amount of bending and its effect depended upon the state of differentiation of the explant at the start of the experiment

Considerable bending occurred in unossified explants from four day embryos with little or no bone formation on the convex (tensile) side but an abnormally large amount on the concave (compressive) side Only very slight bending developed in partially ossified explants from older embryos so that the cartilage was just flattened on one side instead of being convex However, the amount of ossification was always greater on the "convex" than on the concave side

During bending the elasticity of the perichondrium drew it away from the shaft cartilage on the concave side and tension lines radiating from the vertex of the concave side formed in the subperichondral space between the perichondrium and the cartilage Bone deposition followed these tension lines so that the bone at the center of the shaft lay perpendicular to the cartilage The perichondrium on the convex surface was very thin or ruptured so that shallow erosion cavities often appeared In older explants the periosteum was firmly attached to the underlying bone and consequently not drawn away from the concave side of the cartilage In these explants the increased ossification on the convex side followed the tensile stresses which were parallel with the surface of the cartilage

In the second series of studies 77 experiments were made with metatarsals of seven to 12 day embryos and parts of tibias from two day hatched chicks In each experiment two or more skeletal rudiments explanted parallel to each other and a short distance apart, were soon enclosed in a fibrous capsule which later ossified During cultivation the rudiments were gradually drawn together resulting in an alteration of the direction of the tensile stresses in the capsule Two distinct layers were formed in the osseous capsule and during subcultivation the inner one was bent by contraction of the capsule The greatest contraction with

consequent change in stress occurred during subcultivation when the centrifugal stresses in the plasma clot were released so they no longer counteracted the forces pulling the rudiment together Sections of the rudiment showed that the common capsule consisted of successive layers of bone corresponding in number with the changes in the culture medium which were the periods of greatest tension

In the third series of studies 14 experiments using phalanges and metatarsals of ten to 12 day chick embryos were made In these experiments all or part of one epiphysis was removed to change the degree or direction of the normal tensile stresses During growth the epiphysis expanded parallel with and perpendicular to the long axis of the bone shaft thus creating tensile stress in two directions in the periosteum attached distal to the epiphysis Incision of an entire epiphysis removed the lateral component of these forces and greatly reduced the tension while removal of half an epiphysis changed the tensile stresses on just one side of the skeletal rudiment Therefore if tension stimulates osteogenesis either complete or partial incision of an epiphysis should reduce it

This reasoning was verified by experiments in which removal of one epiphysis greatly reduced the amount of bone formed around the shaft After removal of half an epiphysis no bone developed on the operated side of the rudiment although ossification was normal on the intact side Incision of less than half an epiphysis caused no change in the ossification on the intact side However on the operated side bone developed in the superficial periosteal tissue in the area subjected to tensile stress

Glucksman's experiments show that mechanical forces stimulate bone formation and orientation in vitro both during its formation and after differentiation as well as the amount of bone formed in any given region Increasing the normal tension in the periosteum increased ossification while reducing tension decreases it Tensile stress determines where bone will be deposited in osteogenetic tissue In young differentiated bone compressive stresses can change a disoriented structure into a regular pattern Unfortunately Glucksman did not determine the magnitude of the stresses responsible for stimulating osteogenesis or the effect

of excessive stress Presumably an as yet unknown optimum amount of stress is necessary for satisfactory osteogenesis which will not occur if the stress is too little or too great

An experimental approach to the influence of intermittent pressure on osteogenesis in young guinea pigs and rabbits was made by Jores (1920) Bags of water or mercury were taped over the spines of the thoracic vertebrae so that they exerted pressure upon the spinous processes After 100 or more days the animals were sacrificed It was found that under constant unchanging pressure the bone atrophied but active growth occurred on removal of the pressure If the pressure were removed periodically e g, 24 hours on and 24 hours off there was a clear increase in bone growth

Leriche and Policard (1928) explain Jores results on the basis of local calcific changes They believe that osteogenesis occurs under local humoral control pressure leading to a resorption of a certain amount of bone and provoking a local calcium excess Thus pressure is an excitant of bone growth and simply prepares the proper medium for the laying down of bone The latter however, only occurs in the absence of pressure In their opinion mechanical stresses are more important in the preservation and modeling of newly formed bone tissue than in its original formation or development They also review some of the earlier researches on the effects of mechanical influence on bone formation pointing out that, in their opinion, the role of tensile stresses is still debatable Although they accept Roux s idea that shearing stresses cause cartilage formation in connective tissue they do not think it plays any part in osteogenesis

Theories on the Influence of Stress in Osteogenesis

Various theories of the influence of mechanical factors on osteogenesis have been advanced by Thoma and Kokott, discussed by Murray (1936) Weidenreich (1923) and Loeschcke and Weinnoldt (1922) Thoma compared the membranous skull to a closed capsule which increases in size as a result of a uniform internal pressure Assuming that the walls of the capsule are of uniform thickness the stress would be the same throughout the

walls However if the stress were intensified at any point bone would develop in the membranous skull

In order to provide for such an increase in the stress at certain points Thoma postulated that the skull either ceases to grow or fails to react in the usual way to the internal pressure The factor considered responsible for the localized increase in pressure is the bending over of the floor of the skull As development proceeds the outer table of the skull appears as a layer of continuous lamellae, while the inner table is still a network of discontinuous lamellae with bone deposition restricted toward the edges Thoma believes this indicates that the pressure from the brain creates both meridional and latitudinal tensile stresses plus vertical compressive stresses These stresses caused bending of the bone with consequent increased tensile stresses on the outer skull table while they are reduced on the inner table Thus bone deposition is encouraged on the outer and hindered on the inner aspect of the skull except in the peripheral part where the amount of bending is small

Thoma offered no experimental data in support of his views and there seems to be no independent evidence that the bending stresses he mentions even exist The question also arises as to why if Thoma s theory is correct other connective tissue structures e g, the dura and pia mater subjected to the same internal pressure do not ossify Murray mentions a case reported by Weinnoldt, of an individual in whom the brain remained in a single fluid filled vesicle without cerebral hemispheres yet a skull roof with normally arranged bones was present In two cases of anencephali skull bones were recognizable although they were abnormally formed and arranged

The development of the skull bones has also been investigated by Kokott who studied the arrangement of fibrous tissues in the primordial skull of human fetuses two and three months of age Five tracts connecting the areas of the future frontal parietal and occipital bones were identified and considered as the points of anchorage of the skull roof In experiments with an air filled balloon attached at five points corresponding to those in the skull he found that five systems of tension tracts were present and that the balloon closedly resembled the skull roof He, therefore con

cluded that skull form was the result of its five point attachment and an even distribution of internal hydrostatic pressure the five fibrous tracts representing the principal tension lines After attaining its fibrous architecture the skull wall is more resistant to internal pressure along the lines of the fibrous tracts than elsewhere In further development ossification begins in the regions between the fibrous tracts Kokott assumed that the unstrengthened regions would develop a greater stress than the more resistant areas a very doubtful assumption on mechanical grounds

Pressure was considered to be involved in skull osteogenesis by Loeschcke and Weinnoldt (1922) who showed that resorption of the inner skull table occurred with pressure exerted by the skull contents However if the pressure were reduced or absent deposition of bone occurred They believe that pulsations of the brain cause or aid in the circulation of tissue fluids in the bone but are reduced or absent at the points where the brain exerts direct pressure on the skull wall Constant pressure was thought to increase the resistance to the flow of tissue fluids through the bone Removal of the pressure permits the region to have a more efficient vascular system so that more rapid growth results Murray (*loc cit*) points out that this vascular theory cannot apply to rapid alterations of pressure produced by muscular activity or arterial pulsations However two factors provide for relief from constant pressure (1) a compensating vascular apparatus built up during the pressure period, and (2) previously suppressed arterial pulsations transmitted through the cerebrospinal fluid The latter was thought to aid in the flow of tissue fluids

Weidenreich (1923 1924) believed that the mechanical stimulus acting on a bone was mediated through the vascular system However he considered that tension was the type of mechanical stimulus responsible for osteogenesis such as frequently occurs in tendons and other similar structures

Lacroix (1951) believes that the data relating to the influence of mechanical factors on osteogenesis should be divided into two groups The first group consists of the instances in which after a deviation in pressure stresses the bone undergoes a remodeling to better adapt it to the new requirements Examples of this group are genu valgum and genu varum ankylosis or a mal-

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united fracture In these cases the stresses are similar to normal ones but act in a slightly different direction

All other examples he places in a second group characterized by the fact that the stresses acting on the bone are very different in nature and direction from the normal ones The stresses in the first group he considers to be distributed through the articular cartilages and subjacent bone while those of the second group act by compression of the periosteum

The mechanism by means of which mechanical factors induce osteogenesis is unknown Glegg and Leblond (1953) believe pressure causes a dissolution and redeposition of bone and tooth crystals They base their ideas on the principle that a crystal under linear pressure has a greater solubility than one not subjected to such pressure and consequently may dissolve even in a saturated solution Since bone and tooth crystals are surrounded by a fluid apparently saturated with ions composing the crystal the laws governing the growth and dissolution of crystals in a saturated solution should apply to bone They also cite many examples e g erosion of bone by an expanding aneurysm or tumor of the resorption of bone under pressure

Glegg and Leblond believe their concept explains three complex phenomena occurring in bone (1) decrease in intensity of localized radioautographic reactions at the growth sites (2) decrease in the intensity of incremental dentine bands at the dentino enamel junction in young teeth and (3) the decrease in the mineral content of intact bones in animals with fractures These three effects they consider as the direct result of pressure inducing dissolution of bone or tooth crystals followed by the redepositon of part of the dissolved material on less pressed crystals

Summary

The experiments of Glucksman clearly demonstrate that osteogenesis and orientation of trabeculae both during embryonic development and after differentiation are influenced by tensile and compressive forces The same is true of the amount of bone formed in a region Tension determines where bone will be developed in osteogenetic tissue Thus increasing tension in the

periosteum increases ossification, while reducing tension decreases it. In young differentiated bone a disoriented structure can be changed into a regular pattern by compressive forces. If the tensile or compressive stresses are too little or too great osteogenesis will not occur but the optimum amount of stress is unknown and probably varies with bone and species.

Mechanical factors are not entirely responsible for bone form, which also has a hereditary basis as shown by the experiments with chick limb buds in chorio-allantoic grafts. The requirements for maintaining normal shape and proportions of a bone during growth are involved in osteogenesis and the embryonic development of bone. The spatial relations between the periosteum and the diaphysis of a bone are also partly responsible for the increased thickness of the center part of the shaft of the long bone. Mechanical factors are thus seen to be just one of several operating in the development of a normal bone.

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from animals exhibiting any pathology at autopsy were not used The bones were dried from seven to ten days before testing

The breaking load of the fibula of 30 normal normals, the standard controls and the standard group with surgically produced fractures was determined by loading the bone like a beam The breaking load of normal fibulas from rats on the standard diet varied from 205 to 800 gms for the right bone and from 190 to 770 gms for the left one The mean breaking load for the normal right and left fibula was 410 and 458 gms respectively In the other experiments the right fibula was surgically fractured The range of variation in its breaking load was from 0 to 140 gms on the sixth postoperative day the first time the bones were tested, while on the forty fifth day at the termination of the experiments, the variation was from 275 to 650 gms

The influence of a low salt diet on the load which could be successfully supported by the humerus, the radius, and the fibula of young male rats was determined by Clarke Bassin, and Smith (1936) Each experimental group of rats had an age control group of animals fed on an adequate diet and a caloric control group in which the diet although adequate was restricted to the same caloric intake as in the low salt group

In the low salt experiments some of the rats were kept on the diet for three weeks and others for six weeks During this time there was a drop in the load supported by bones from rats on the low salt diet Thus the load the humerus supported decreased from 1586 to 1195 gms the radius from 624 to 429 gms and the fibula from 149 to 135 gms The load borne by the humerus in the age control group rose from 3139 to 4279 gms of the radius from 1313 to 1621 gms and the fibula from 264 to 294 gms In the "caloric control group the successfully supported load increased from 3106 to 3659 gms for the humerus from 1187 to 1262 gms for the radius and from 234 to 332 gms for the fibula

Realimentation experiments were then performed in which rats after being on a low salt diet for three and six weeks were divided into three groups the first of which was placed on a normal diet for an additional nine weeks the second group for an additional six weeks and the third for an additional 12 weeks In the first group the fracturing load for the humerus increased

13

Factors Influencing the Breaking Strength of Bones

THE EFFECT of diet disuse, and other factors on the breaking *strength or stress in bones of experimental animals has been* studied by several investigators Most of the studies show that the breaking stress of bones is definitely affected by nutrition, hormones immobilization of the bone and certain poisons The results of some of the more recent investigations of this nature are discussed in the present chapter

The Influence of Diet

McKeown Lindsay Harvey and Howes (1932) studied the effect of diet on the load which healing fractures of the fibula could support before breaking Rats of a known stock four to eight months of age and weighing between 190 and 300 gms were used The rats were fed on a standard Moise and Smith diet for a week and then were divided into three groups The 50 rats of the first (normal normals) group were sacrificed after one week on the diet and the breaking load (gms) of the normal right and left fibulas was determined The second (standard control) group also fracture free was continued on the diet for a longer length of time At the start of the experiment the rats were divided into lots of four and from the thirteenth to the fifty second day one lot was sacrificed at three day intervals The third (standard) group consisted of lots of seven rats each of which had been on a diet for a week at the end of which the right fibula was surgically fractured by cutting it with scissors At 14 different times during the experiment the breaking load of the fibula was determined on four standard control animals without fractures and seven standard animals with fractures Bones

The influence of calcium intake on the breaking strength and size of the femur was studied by Bell Cuthbertson and Orr (1941) in 96 recently weaned male albino rats Four experiments were made on groups consisting of four or five rats weighing about 60 gms One representative group was sacrificed at the beginning of each experiment so that the calcium content could be determined by weighing the eviscerated carcasses The remaining groups for a period of 56 days were placed on diets of constant protein carbohydrate fat and vitamin content but varying in amount of calcium The authors were aware of the variation in the Ca P ratio in their experiment but presumed that any rachitogenic action resulting from the abnormal Ca P ratio was prevented by giving adequate amounts of vitamin D

At the end of an experiment the femurs were cleaned, x rayed weighed and measured After the mechanical tests the calcium content of the bone was determined Before testing the bones were dried at room temperature for several days to prevent weight loss during the tests The breaking strength of the right femurs was determined by bending while the left femurs were tested by torsion

For the bending tests the femur was supported horizontally at the ends and loaded to rupture at the middle of the shaft The actual fracturing load was taken to lie between the greatest weight supported and the final weight applied Two micrometer measurements taken perpendicular to each other, were made of the thickness of the cortex at the fracture site During the tests slight variation in the length of the span was allowed The greatest bending moment was used as the strength index of the bone The deflection of the midpoint of the shaft during a test was measured by a dial gage resting lightly on the wire supporting the pan for the weights

For the torsion tests the ends of the bones were cemented in hollow brass cubes One end of the bone was fixed while the other was twisted by placing weights on a pan suspended from a lever attached to the bone perpendicular to its long axis Weights were gradually added to the pan until the bone broke Generally the free end of the bone rotated through an angle of about 30 before it broke The twisting strength was measured

from 1536 to 5042 gms for the radius from 624 to 1897 gms, and for the fibula from 149 to 517 gms

The breaking load borne by the bones of rats of the third group showed a definite increase over the strength of bones from rats of the second group The breaking load for the humerus rose from 3968 to 4280 gms for the radius from 1675 to 1952 gms, and for the fibula from 374 to 387 gms In the age control rats the breaking load for the humerus increased from 3761 to 5189 gms for the radius from 1604 to 1909 gms, and for the fibula from 306 to 517 gms

A few studies of the breaking load of leg bones of cattle on various diets have also been made Becker and Neal (1930) found that the leg bones of a cow on a low calcium ratio broke under a load of 335 lbs The bones from three cows on a diet containing 2% bonemeal required 2345 to 3450 5 lbs to fracture them while the bones from five Angus cattle did not break until loads of 2713 5 to 3149 0 lbs had been applied The bones from cows on a diet deficient in P and $CaCO_3$ broke with loads of 2311 5 to 2680 lbs while those from a cow that had fully recovered from P deficiency did not break until 3618 lbs were applied In another paper Becker Neal and Shealy (1934) found that the breaking load of 215 bones from Florida cattle on a calcium deficient roughage ranged from 335 lbs in a cow that suffered a fractured pelvis to over 3000 lbs in seven cows that received bonemeal for 13 to 27 months

In the above discussed papers the authors constantly refer to the breaking strength of the bones although they never actually determined it The breaking strength or stress of a material is computed in terms of the load (pounds or kilograms) per unit area (square inches or square millimeters) which the material supports up to the rupture point In order to compute the breaking strength the cross section area of the specimen of material tested must be known None of the above mentioned authors determined the cross section area of the bones they tested Therefore the breaking strength or stress of the bones cannot be calculated Clarke Bassin and Smith gave the diameter of the bones they tested but the cross section area of the bones cannot be computed from this data because the bones are hollow

soon as possible after sacrifice while another was not tested for two weeks and a third not for ten months after removal and cleaning Air drying of a bone produced no significant alteration in its strength In both cases the difference in bending moment between the fresh and the air dried bone was less than 2%

In analyzing their results it was found that the strength of bones, indicated by the bending and torsion moments increased with increased calcium intake up to about 0 36% Ca in the diet With higher calcium intake there was no further increase in bone strength The average breaking stresses in bending and twisting were approximately 35000 and 9500 lbs/in², respectively The ratio of breaking stress in twisting to that in bending (0 27) is of some interest For isotropic material, e g cast iron and steel this ratio is slightly more than half but for laminated material like wood the ratio is considerably less about 0 1 This is considered to be due to the weakness of cement substance between adjacent lignified fibers Evidently bone has a somewhat similar weakness at the interconnections between the longitudinal fibers In the few cases in which it was investigated the load deflection curve was a straight line to failure

In a second paper Bell Chambers and Dawson (1947) tested the effect of a rachitogenic diet upon the strength and other physical properties of the femur of male albino rats about four to eight weeks of age After weaning at three weeks the animals were fed a full diet until they reached a weight of about 50 gms One of each litter was killed so the carcass after removal of the stomach and intestines could be analyzed for Ca and P One of the remaining litter mates was put on a rachitogenic diet supplemented with vitamin D (group N), while one or two of the other animals were put on the rachitogenic diet without the vitamin D (group R) A fourth animal from each litter (group S) was fed on a standard diet The methods for measuring breaking stress on bending and twisting were the same as described by Bell Cuthbertson and Orr (*loc cit*)

One of the most interesting results of their experiments was the paralysis of the hind legs of the rats in group R The paralysis began three weeks after the start of Experiment I and progressed until it was almost complete in some animals No paralysis was

by the product of the fracturing load and the leverage (weight times length of lever) The thickness of the cortex was measured at several places on the fracture, which often ran obliquely or spirally along the shaft for some distance

In the torsion tests the load should have been applied to the bone at the periphery of a wheel instead of by a lever because with a 30 angle of torsion the length of the lever arm would change appreciably Consequently the effect of the load upon the bone would also change during a test

In contrast to the previously discussed dietary studies the breaking stress of the bone was calculated to determine if lack of calcium in the diet affected the quality of the bone formed under these circumstances Assuming that the section profiles of the various specimens were similar in shape and proportional dimensions the breaking stress was calculated from the formula $M = s_1(abt)$ and $T = s_2(abt)$ where M equals bending and T equal twisting moments at fracture s_1 and s_2 equal breaking stress in bending and twisting respectively, and *a* *b*, and *t* are the dimensions of the central or fracture section i e, *a* is the outside diameter in one direction *b* the outside diameter in the direction perpendicular to the first and *t* the thickness of the cortex These formulae do not give absolute values for breaking stress but values comparable with one another

Since the section profiles are not exactly similar the error was checked by use of the formulae $M = s_1 \frac{\pi}{32}\left(\frac{ab^3 - dc^3}{b}\right)$ and $T = s_2 \frac{1}{2} \pi (a - t)(b - t)t_m$ where *a* and *b* are the outside diameters taken perpendicular to each other and *d* and *c* are the corresponding inside diameters and t_m is the minimal thickness of the cortex The latter two formulae are based on the assumption that the sections are elliptical in shape and composed of perfectly elastic material These latter formulae provide absolute values for breaking stresses s_1 and s_2 In using these formulae it is assumed that the stress strain curve is a straight line to failure and that the spongy bone has no strength

The effect of drying on the strength of bone was tested by comparing femurs from two animals One femur was tested as

soon as possible after sacrifice while another was not tested for two weeks, and a third not for ten months after removal and cleaning Air drying of a bone produced no significant alteration in its strength In both cases the difference in bending moment between the fresh and the air dried bone was less than 2%

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present in the rats of group N In Experiment II an unsuccessful attempt was made to eliminate the paralysis by adding vitamin E to the diet A further attempt to avoid paralysis was made (Experiment III) by adding vitamins E and A to the diet In group R five rats had severe paralysis at the end of the experiments, four others showed weakness, and only two had no evidence of it at all

The breaking stress according to group for each of the three experiments is summarized as follows

Experiment	*Groups* R	N	S
I	11710 lbs /in	18700 lbs /in	————
II	12850 lbs /in	18130 lbs /in	————
III	11860 lbs /in	18590 lbs /in	26890 lbs /in

A high correlation between the bone ash content (%) and the breaking stress was indicated by the correlation coefficient *r* which was +0 52 (S E = 0 15) for bending and +0 405 (S E = 0 12) for twisting The structure of the bones, as revealed by their x ray diffraction patterns was also investigated However the authors concluded that the mechanical chemical and x ray findings indicated no disturbance in the fundamental plan of ossification in rachitogenic rats

In a third paper the strength and elasticity of bones of rats on a rachitogenic diet was studied by Weir, Bell and Chambers (1949) The bones used were obtained during the experiments of Bell Chambers and Dawson (*loc cit*) just described The bending tests were made by the method described by Bell Cuthbertson, and Orr (*loc cit*) Twelve animals were on the rachitogenic diet (R group) 19 on the rachitogenic diet supplemented with vitamin D (N group) and six animals on a complete diet (S group) In contrast to the earlier papers of this series the mean strain in the bones at the elastic limit and at rupture was also determined (Table III)

Stress strain diagrams were drawn which showed that while stress at the elastic limit was greater in bones having the higher ash content strain was very similar in all the bones Bones with a higher ash content also had greater stress at the breaking point although a tendency was noted for the strain at rupture to be

TABLE III

STRESS AND STRAIN VALUES FOR RAT FEMURS

(Data from Weir, Bell and Chambers 1949)

	Group		
	R	N	S
Mean strain at elastic limit (%)	1.27	1.28	1.30
Mean strain at rupture (%)	2.11	2.18	1.84
Mean strain at elastic limit as % of mean strain at rupture	60	59	71
Mean stress at elastic limit (lb/in²)	8100	12600	21200
Mean stress at rupture (lb/in²)	11900	18900	26900
Mean stress at elastic limit as % of mean stress at rupture	70	67	79

higher in bones from rats on a poor diet when the ash percentage was low. The differences which diet produced in the breaking stress on bending, in the modulus of elasticity, and in the percentage of ash in rats of the different groups, was highly significant as revealed by Student's *t* test.

Plotting breaking stress and the modulus of elasticity against the percentage ash suggested a positive correlation between them. These are intrinsic values independent of bone size. In the bones from groups N and S a positive straight line relation was found when breaking stress was plotted against the modulus of elasticity. A highly significant relation between these two physical properties of bone, as indicated by a correlation coefficient of + 0.63, was found in all three groups, R, N, and S, even after elimination of the effects of diet. The results of the experiments indicate that breaking stress and the modulus of elasticity appear to be more closely related to each other than either is to the percentage ash in the bone.

Although the stress at the upper limit of elasticity had a wide range of variation in the three groups of rats, the strain at the same point was very constant at about 1.5%. The authors believe that the modulus of elasticity, because of its close association with the breaking stress on bending, is a good index of the quality of bone and might be used in predicting the load a bone can safely bear. In spite of the significant differences in the breaking stress of bones from rats fed on the different diets they are not convinced that higher ash content alone is responsible for the greater stress and modulus values in bones from animals fed on the better diets. This conclusion was based on the Ca : P ratio

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true of the femurs in the thyroid experiments. However, the difference in the quality of the bone, as indicated by the breaking stress (Table IV), of the experimental and control animals was slight.

TABLE IV

ABSOLUTE BREAKING STRESS IN BONES OF HORMONE TREATED RATS

(Data from Bell & Cuthbertson, 1943)

Group I	
Ox anterior pituitary	30000 lbs/in²
Calf thymus controls	3[illegible]0 lbs/in²
Group II	
Horse anterior pituitary	31500 lbs/in²
Sheep anterior pituitary	35800 lbs/in²
Ox anterior pituitary	36100 lbs/in²
Calf thymus controls	36300 lbs/in²
Group III	
Ox anterior pituitary	33530 lbs/in²
Calf thymus control	37360 lbs/in²
Oestradiol dipropionate	34270 lbs/in²
Parathyroid hormone	32670 lbs/in²
Controls	31620 lbs/in²
Group I	
Thyroid	32530 lbs/in
Controls	34300 lbs/in²
Group II	
Thyroid	31300 lbs/in²
Controls	28000 lbs/in²

Bell and Cuthbertson concluded that the differences in breaking strength of the bones of the experimental vs. the control animals were the result of changes in the dimensions of the bones rather than in the quality of the bones. The latter characteristic, as indicated by the slight differences in the breaking stress of the bones, was not changed by the hormones given the animals during the experiments.

The Effect of Disuse

The effect of disuse on the breaking strength, extreme fiber stress, gross and microscopic anatomy, chemical composition, and permeability to x rays of foreleg bones in dogs was investigated by Allison and Brooks (1921). In 13 experiments the foreleg

which was two in all bones and the similarity in the structure of the inorganic matrix as revealed by x ray crystallography They feel that it is safer to assume that alterations in the physical properties of bone are the result of variations in the relative proportions of the organic and inorganic constituents

In an addendum to the paper Walter L M Perry pointed out that there was no significant correlation between ash and breaking stress or between ash and the modulus of elasticity in any one separate group Also the relation between breaking stress and modulus was not significant for bones in group R but was highly significant in bones from the other groups Since diet might affect the quality of collagen as well as the ash content, so that breaking stress and elasticity could vary independent of the ash content he thought it desirable to combine the data from the three groups without eliminating the effect of differences in diet When this was done a highly significant correlation was found for ash vs elasticity ash vs breaking stress, and for breaking stress vs elasticity

The Influence of Hormones

The influence of hormones on the breaking strength and stress of white male rat femurs was studied by Bell and Cuthbertson (1943) The animals were divided into three groups, I II and III and given ox horse and sheep interior pituitary extract oestradiol dipropionate, parathyroid and thyroid hormones The thyroid hormone was given with the food but the others were injected subcutaneously The controls of the same age received no treatment except in the pituitary experiments in which the control animals were given extract of calf thymus instead of pituitary extract

The right femurs were tested for breaking stress on bending and the left ones for breaking stress by twisting The methods of testing and computing stress were the same as used by Bell Cuthbertson, and Orr (*loc cit*)

The results of the experiments showed that while the bones from animals treated with extracts of the interior pituitary oestradiol dipropionate and parathyroid were proportionately larger and stronger than those from the controls the reverse was

were cut. In a third group of eight kittens anterior root section was combined with removal on the side of the operated limb of all the lumbar sympathetic ganglia from the renal vein to the pelvic brim. In all experiments the opposite, normal hind leg served as the control.

Two months postoperatively the kittens were sacrificed. The leg bones were then cleaned, weighed and exposed to air drying at room temperature for several weeks so that the moisture content of all the bones would be approximately equal.

Additional experiments were made on 16 young white, male rats weighing approximately 120 gms. One hind leg was paralyzed by avulsion of the femoral and sciatic nerves. The animals were sacrificed four weeks postoperatively and the femurs and tibias removed, cleaned and dried.

The kitten and rat femurs were then subjected to bending tests in order to determine the bending moment, the breaking stress and the modulus of elasticity. The method of testing was the same as used by Bell, Cuthbertson and Orr (1941).

The results of the tests (Table V) showed that the average ultimate bending stress in the kitten bones was greater in the bone from the denervated limb in the animals in which the anterior (Group 1) and the posterior (Group 2) roots of the spinal

TABLE V

ULTIMATE BENDING STRESS (LBS/IN²) IN THE LONG BONES OF THE HIND LIMB OF KITTENS AND RATS

(Data from Gillespie 1954)

Animal	*Average Ultimate Stress* Denervated	Normal	*Range of Variation* Denervated	Normal
Kitten				
Group 1	38960	37960	34100–43300	35800–40000
Group 2	38800	37933	31100–44900	32000–42900
Group [illegible]	38187	44850	31300–41800	41300–47600
Rat	27092	31835	18400–33300	25200–35900

Animal	*Stress Greater on*	*% Difference*	*Statistical Significance*
Kitten			
Group 1	Denervated Side	2.8	Not significant
Group 2	Denervated Side	2.3	Not significant
Group [illegible]	Normal Side	14.9	Significant
Rat	Normal Side	14.4	Significant

was partially or completely paralyzed by sectioning of the brachial plexus, in seven experiments a flail joint was produced by incision of the proximal end of the humerus, and in four experiments the foreleg was fixed in plaster of Paris The length of an experiment varied from a few days to almost a year (314 days) At the completion of an experiment the bones from the used and nonused limbs were compared by x ray examination, measurement, weights chemical composition, breaking strength and extreme fiber stress

The breaking strength was determined by placing the bone on two supports 25 mm apart, and loading it to failure at the mid point between the supports Whole metacarpal bones from the used and nonused limb of an adult dog in which the left fore leg had been paralyzed for 113 days by section of the brachial plexus were compared The weight necessary to break the second third fourth and fifth metacarpals was 21200 24000 21300 and 18000 gms respectively, for the used limb, and 7100, 7000 6000 and 9000 gms respectively for the nonused limb

In order to determine whether the decreased strength of the intact bones from the paralyzed limb was because of reduction in the form and dimensions of the bones or because of intrinsic changes in the bone matrix rectangular pieces of bone from the humeral cortex were tested The width and thickness of the pieces at the fracture point were measured with micrometer calipers and the extreme fiber stress computed by substituting these values in the formula used for calculating similar stress in rectangular beams supported at the ends and loaded in the middle Little difference was found in the extreme fiber stress from the two types of bone the piece from the used humerus having a stress of 12000 gms/mm^2 while that from the nonused bone was 15000 gms/mm

Another study of the effects of disuse on bones was made by Gillespie (1954) who tested some of the physical properties of femurs tibias and fibulas from kittens and white rats Twenty eight kittens, weighing from 750 to 1200 gms were used In the first group of ten kittens the anterior roots of the spinal nerves innervating one hind limb were sectioned In a second group of ten kittens the posterior roots of the corresponding spinal nerves

metatarsals by torsion The methods of testing were the same as employed by Bell Cuthbertson and Orr (1941)

The average bending moment of the fluorotic bones was 25% greater than that of the control bone but the results were not directly comparable because the bones came from sheep of different ages and breeds These difficulties did not apply to the values for breaking strength by bending and by torsion because these factors are independent of the size and shape of the bone

The breaking or ultimate strength on bending varied from 24400 to 26300 lbs/in² in the fluorotic femurs, from 28100 to 32200 lbs/in² in the fluorotic metacarpals, and from 26600 to 31311 lbs/in² in the fluorotic metatarsals The range of variation in the control bones was 16700 to 27000 lbs/in² for the femurs 20100 to 33000 lbs/in² for the metacarpals and 26000 to 37400 lbs/in² for the metatarsals The average breaking strength by bending was 29000 lbs/in for the affected bones and 27300 for the control bones

The range of variation for breaking strength in torsion was less than that for bending Among the fluorotic bones the strength varied from 8000 to 12000 lbs/in while the control bones had a variation of 11800 to 14500 lbs/in² The average breaking strength for the fluorotic and control bones was 10400 and 13000 lbs/in², respectively

The fluorotic bones were generally bigger heavier and thicker in proportion to length than were the control bones No appreciable difference from the control bones was found in the ash content of the fluorotic femur, the actual values being 69 1% and 73 9% respectively Crystallographic studies by x ray of two fluorotic and one control bone revealed no differences in the spacing or orientation of the spectra obtained The authors point out that on theoretical grounds such disturbances in the spectra would be unlikely because of the small amount (1%) of fluorine in the affected bones The mean value for fluorine was 1 02% of the ash in the affected bones and 0 028% in the control bones

Bell and Weir attempted to compute the bending moment and the breaking strength on bending in bones from pigs from the data given by Kick Bethke and Edgington (1933), who fed pure sodium fluoride or rock phosphate containing fluorine to

nerves to the limb had been sectioned In the kittens in which the sectioning of the anterior spinal nerve roots was combined with a lumbar sympathectomy (Group 3) the bones from the normal limb had the greater average ultimate bending stress This was a statistically significant difference while that in groups 1 and 2 was not In the rats the average ultimate stress in the bones from the normal limb was also statistically significantly greater than that in the bones from the denervated limb In all bones from the kittens the average ultimate stress was considerably greater than that in the rat bones

Gillespie concluded that the altered changes in the physical properties including ultimate stress of the bones from the paralyzed limbs was almost entirely due to decrease in the quantity of bone resulting from secondary loss of muscle activity There was no evidence that vascular changes were involved or that nerves exert any specific trophic influence on bone

The Influence of Healing

The breaking strength of healing fibular fractures in young healthy adult male rats has been measured by Lindsay and Howes (1931) The fractures were surgically produced by cutting the fibula with scissors The cleaned bones were placed in a desiccator for 24 hours before testing and then the breaking load determined by loading the bone like a beam The average load required to break normal nonfractured fibulas varied from 285 to 580 gms The experiments with the fractured fibulas lasted 45 days at which time the bones were completely healed The average load necessary to break these bones was 85 gms at the sixth postoperative day 341 gms at the twenty first postoperative day 285 gms at the thirtieth postoperative day and 465 gms at the forty fifth postoperative day

The Effect of Fluorosis

The breaking strength under bending and torsion of long bones from four sheep with fluorosis and from four normal animals was determined by Bell and Weir (1949) Right femurs metacarpals and metatarsals were tested by bending and left

with diets low in salt calcium and phosphorus The breaking strength of rat bones as indicated by the bending and torsion moments, increases with increased calcium intake up to about 0 367 Ca in the diet but greater Ca intake has no further effect on bone strength The breaking strength of rat bones is decreased by a rachitogenic diet although the strength can be increased somewhat if the diet is supplemented with vitamin D However the strength cannot be brought up to that of bones of animals fed on a standard diet Bones with a higher ash content have a greater stress at the elastic limit and breaking point, although the strain is very similar in all bones A tendency has been noted for bones from rats on a poor diet with a low ash content to have a higher strain at failure

Giving rats extracts of anterior pituitary oestradiol dipropionate and parathyroid produced proportionately larger and stronger bones than those of the controls but the reverse was true of the femur of rats given thyroid hormone with their food However the difference in the breaking stress of the bones from the experimental and control animals was slight

No statistically significant differences were found between the ultimate breaking strength of bones of kittens in which the anterior and the posterior roots of the spinal nerves had been sectioned and bones from the normal control limbs However, the breaking strength of bones from kittens in which section of the anterior spinal nerve roots was combined with lumbar sympathectomy was significantly less than that of the normal control bones The decrease in the strength of the bones from the paralyzed limbs is probably the result of decrease in quantity of bone from secondary loss of muscle activity rather than from vascular changes or trophic influence of nerves

Fluorine apparently reduces the breaking strength by bending and torsion of sheep and pig bones However the evidence is not entirely clear and the mechanism of the action of fluorine is not well understood

112 pigs. These animals had up to 1.108% of fluorine in dry fat free bone, which is equivalent to 1.91% of ash. The bending strength was computed from the formula $\frac{32\,M}{\pi\,\delta^3}$ where δ is the smallest external diameter of the bone. However, the values obtained are meaningless because the formula is for a solid circular area rather than a ring shaped section such as from a shaft of a bone.

With an increase in the percentage of fluorine there was a steady decrease in the breaking strength by bending in the pig and sheep bones. In a previous investigation Bell, Chambers and Dawson (*loc. cit.*) found that the breaking strength of rat bones decreased when the percentage of ash was greatly reduced but the ash content was nearly the same in all the pig bones. Even in bones with a high fluorine content the reduction in ash content was too small to account for the decrease in breaking strength. Bell and Weir, therefore, concluded that the fluorine rather than the ash content was responsible for the reduction in breaking strength of the bones.

They believe the fact that the bending moment remained high until a considerable amount of fluorine was present, while the breaking strength declined steadily, indicated a compensatory mechanism was present. This compensation could be effected either by an increase in the thickness of the cortex or an increase in the thickness of the shaft. Kick *et al.*, Phillips *et al.*, Peirce and Bosworth *et al.*, cited by Bell and Weir (*loc. cit.*) agree that fluorotic bones from different animals all exhibit an abnormal increase in thickness of the cortex and shaft. The affected sheep bones were thicker than the control bones. The same was true in the pig bones. Even in severely fluorotic bones, in which compensation was probably incomplete, the breaking strength of the bone was reduced about one third. The pig femur with the greatest fluorine content required approximately 400 lbs. to fracture it, while the weakest sheep femur supported over 500 lbs. before breaking.

Summary

Experimental studies with rat and cattle bones show that the breaking load, when the bone is tested like a beam, is decreased

changed into kilograms per square millimeter by multiplying by 0.000703.

Tensile Stress and Strain

The first significant experimental investigation of the tensile strength of human bone was made by Wertheim (1847) who attempted to determine the strength of intact human fibulas under direct tension. However, he had no satisfactory method of holding the bone and found that a load of 140 kg. pulled the condyles away from the compacta with only slight elongation or strain in the shaft. He, therefore, tested long thin straight strips of bone from the compacta of femurs and fibulas of fresh cadavers of individuals from one to 74 years of age. Each specimen was subjected to tensile loads up to failure and its strain determined by measuring, with a cathetometer, the distance between two reference marks on the specimen. Measurements were taken with and without load. The cross section area (mm^2), load (kg/mm^2), length of the specimen with and without load, and the observed and calculated elongation (for one meter of length), as well as other physical properties were recorded. Data were reported from eight samples of the femur and fibula of two females, 21 and 60 years of age, and from two males, 30 and 74 years of age. The specimens were tested fresh, with the load applied parallel with the long axis of the specimen and the intact bone. The testing apparatus was not illustrated nor its accuracy stated.

The ultimate tensile strength in the samples from the female femurs varied from 9100 lbs/in^2 in the 60 year old individual to 9770 lbs/in^2 in the individual 21 years of age. The strength of the fibular samples of the same individuals was 4960 lbs/in^2 and 14590 lbs/in^2, respectively. The samples from the male femurs had a strength of 10380 lbs/in^2 for the 60 year old man and 14930 lbs/in^2 for the one 30 years of age. The fibulas of the same individuals had a strength of 6160 lbs/in^2 and 21370 lbs/in^2, respectively. The average strength in the samples from the male and female individuals, respectively, was 13210 and 9538 lbs/in^2. No consistent decrease in tensile strength with advancing age of the individuals was noted.

One of the most extensive studies on the physical properties of bone was made by Rauber (1876) who determined the tensile

14

The Tensile, Compressive and Shearing Strength of Bone

In contrast to the preceding chapters in which stress and strain in intact bones were discussed the present and following chapter deal with the same phenomena in bony tissue These phenomena are studied by preparing samples or specimens of standardized size and determining their stress strain characteristics by the same methods employed by engineers for evaluating similar properties of structural materials Factors influencing the stress strain values obtained are the amount of moisture in the sample whether the sample was from embalmed or unembalmed, compact or spongy bone the shape and cross section area of the sample the direction of application of the force with respect to the axis of the bone from which the sample was obtained or the axis of the collagen fibers of the sample, and biological factors such as the age state of health and possibly sex and race of the individual from whom the samples were taken

The strain and the ultimate strength of the samples are the physical properties usually determined by a test the ultimate unit stress being computed later However when the sample is tested to failure the ultimate unit strength i e the load per unit area at which failure occurred is equivalent to the ultimate unit stress In conformity with common engineering practice the term strength will be used in the following discussion instead of stress although the two may be considered as synonymous Most of the studies on stress and strain in bony tissue have been made by Europeans who report their stress values in terms of kilograms per square millimeter English and American workers compute stress in terms of pounds per square inch Kilograms per square millimeter can be converted into pounds per square inch by multiplying by 1422 Pounds per square inch can be

changed into kilograms per square millimeter by multiplying by 0.000703.

Tensile Stress and Strain

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One of the most extensive studies on the physical properties of bone was made by Rauber (1876) who determined the tensile,

compressive, shearing bending and torsion strength of samples of compact bone from the middle of the shaft of the human humerus, femur, and tibia, as well as tibias from an ox a calf, a domestic hog and a wild hog Additional tests were made on samples of spongy bone from human lumbar vertebrae a femur, and a tibia The human material was obtained from male individuals 28, 30 33, 40, and 70 years of age and from one female 56 years of age

Most of the samples were tested in the fresh condition after being warmed in water at a temperature of 38 C, but a few, dried at a room temperature of 15 - 25 C, were also studied The ultimate tensile strength was determined for 30 femoral eight tibial and 11 humeral samples The samples were tested under direct tension with the direction of the force parallel with the long axis of the intact bone and the collagen fibers The samples had a central reduced area 3 cm long with a cross section area of 8 mm²

Rauber recorded the fracturing load for each sample and the average ultimate tensile strength (kg/mm) for all the samples from each bone However he did not compute the ultimate tensile strength for each sample This has been done by the author, using Rauber s data for fracturing load in order to determine the range of variation in the ultimate strength among the samples from a single bone as well as among samples from different bones

The average ultimate tensile strength and range of variation of the human material according to bone, was 14560 lbs/in (8880 18670) for the humeral samples 17640 lbs/in (15110 24170) for the tibial samples and 12850 lbs/in² (8170 18840) for the femoral samples The humeral samples from the man 30 years of age had the greatest average strength (15780 lbs/in²), while those from the man 28 years of age had the least (13150 lbs/in) The tibial samples were all from the man 30 years of age The femoral samples with the highest average strength (15860 lbs/in) were from the same man while the weakest (9020 lbs/in²) were from the woman 56 years of age The six dry femoral samples from the man 28 years of age had an average strength of 16460 lbs/in² (14220 18670)

Another extensive investigation of the strength and other

physical properties of bone was Hulsens (1896), who also determined the ultimate tensile strength of his samples under direct tension in their long axis. Eight fresh samples (four from the humerus and four from the tibia) of an adult man and ten dry samples (four from the humerus of the same man and six from the femur of another) were tested. Additional tests were made on samples from the femur of an ox, a calf, and a wolf, and a tibia of an ox. All these samples were cut parallel with the long axis of the intact bone but five others, cut transverse to the long axis, were taken from a fresh ox tibia. Six decalcified samples, three parallel with and three transverse to the long axis of an ox tibia were tested. The samples had a reduced center section with a cross section area of 6 or 9 mm². Tensile failure generally occurred at a point ⅓ the length of the sample and the decalcified samples split into slivers. An Amsler Testing Machine of unstated accuracy was used.

The average tensile strength for the human material was 15000 lbs/in² (13380–16280) for the fresh humeral samples and 15550 lbs/in² (14850–17480) for the dry samples from the same bone. The fresh tibial samples had an average tensile strength of 15040 lbs/in² (12880–16250) and the dry femoral samples an average of 16850 lbs/in² (14260–21720).

The ultimate tensile strength of 11 samples from the femoral shaft of ten embalmed cadavers was determined by Carothers, Smith and Calabrisi (1949). The average age of the individuals whose bones were tested was 51 (30–75) but the age and sex of the individuals, according to samples, were not recorded. There was an equal number of whites and Negroes but five times as many men as women. None of the individuals died from musculoskeletal disease.

The samples had a cross section area varying from 0.0179 to 0.0194 in² and were tested to failure under tension in the long axis of the fibers. A 600 lb capacity hand operated Amsler Testing Machine was used. The loading speed was about 100 lbs/minute. The ultimate tensile strength varied from 16600 to 31500 lbs/in² with an average of 22000 lbs/in².

In a more extensive study Evans and Lebow (1951) determined the ultimate tensile strength and per cent elongation

(strain) under tension, as well as other physical properties of 242 samples of compact bone from the shaft of 12 femurs (6 pairs) An attempt to determine regional variation in the physical properties within a single bone was made by cutting samples of a standardized size from the anterior, posterior medial, and lateral quadrants of the proximal middle, and distal thirds of the shaft The influence of moisture on the physical properties was investigated in each bone by a comparative study of the same physical properties in wet and dry samples from each region of the bone One sample from each region was air dried at room temperature and tested dry, while the immediately adjacent sample was kept in water and tested wet All samples had a reduced area 1 25 inches long and a cross section area of 0 0135 in The femurs used in the study came from embalmed bodies of white males varying from 47 to 81 years of age None of them had died from a primary bone disease

The ultimate tensile strength (stress) of the samples was determined under direct tension in the long axis by loading them to failure in a 5000 lb capacity Riehle Testing Machine calibrated to an accuracy of ± 1% The low range scale of the machine (0 250 lbs) was used so that the load applied was registered in units of 0 5 lbs

The strain occurring in a sample during a test was measured, in units of 0 0001 inches by a Porter Lipp extensometer over a gage length of one inch The magnitude of the strain was reported in percentage elongation which the sample underwent during a test If the fracture occurred outside of the gage marks of the extensometer the test was not used

The results of the tests showed that air drying increased the average ultimate tensile strength of the samples but decreased the strain or percentage elongation The wet samples from the middle third of the shaft were the strongest (12070 lbs/in) while those from the proximal third were the weakest (11260 lbs/in) The wet samples from the middle third of the femoral shaft also exhibited the greatest strain or elongation (1 27%) while the samples from the distal third had the least (1 15%) With respect to quadrants the wet samples from the lateral quadrant were the strongest (12120 lbs/in²) while those from the

anterior quadrant were the weakest (11370 lbs/in²) The greatest percentage elongation occurred in the wet samples from the medial (1 35%) and lateral (1 34%) quadrants while the posterior quadrant samples had the least elongation (1 06%) The differences between the maximum and minimum values were 7% for ultimate tensile strength and 10% for strain, considered by thirds, and 12% considered by quadrants The average ultimate tensile strength for all wet samples, regardless of region, was 11840 lbs/in² while the strain or elongation was 1 20%

The authors considered the results from the wet samples were the more significant and a closer approximation to the true values for the tensile strength and strain of living bone An attempt was made to determine the moisture content of compact bone by oven drying at a temperature of 105 C, all the samples from the femur of a white male 78 years of age When a constant weight was obtained after prolonged drying it was found that

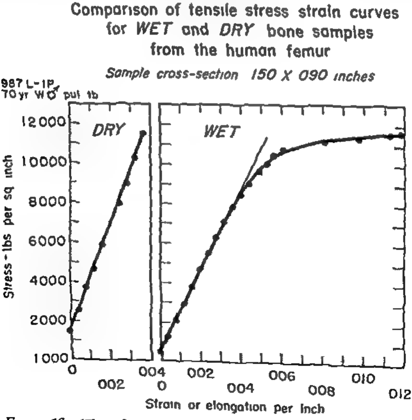

Figure 38 (From Evans and Lebow *J Appl Physiol* 3 1951)

the samples had decreased 12% in weight because of moisture loss

As far as the author is aware, Evans and Lebow (*loc cit*) are the only investigators who have studied the influence of moisture upon the energy absorbing capacity of bone Stress strain curves (Figure 38) of dry and wet samples of standardized size from the same region of a single bone clearly demonstrated the greater energy absorbing capacity of the wet sample as evidenced by the greater size of the shaded area beneath its stress strain curve The magnitude of the energy (in lbs/in^3) each sample absorbed to failure can be determined by measuring with a planimeter the shaded area below the stress strain curve The total energy absorbed to failure by all the wet samples from the femur of a white male 78 years of age amounted to 83 6 in lbs/in^3 while the dry samples from the same bone absorbed only 69 8 in lbs/in^3

The same investigators were also the first to demonstrate that the strain or percentage elongation occurring in a sample under direct tension is directly proportional to its energy absorbing capacity This was shown by the straight line obtained (Figure 39) when the percentage elongation of wet and dry samples from the left femur of a white male 78 years of age were plotted against the energy (in lbs/in^3) they absorbed to failure The less steep slope of the line for the wet samples indicates their greater energy absorbing capacity

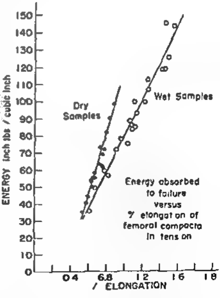

Figure 39 (From Evans and Lebow J Appl Physiol 3 1951)

According to Best and Taylor (1950) water constitutes about 25% of bone weight thus increasing the energy absorbing capacity of living bone The greater energy absorbing capacity of wet bone is of extreme importance in traumatic fractures most of

which arise from blows or impacts and are energy problems Consequently, the ability of a bone to absorb a considerable amount of energy and to undergo marked deformation without breaking are safety factors

In a second paper Evans and Lebow (1952) used the same methods in studying variations in the ultimate tensile strength of the femur, tibia and fibula from the same individuals The bones were obtained from the embalmed body of a white male 36 years of age another 47 years of age and a Negro male 72 years of age They were selected as representing a young middle aged and an old man The bones from both legs were used The total number of samples tested was 115 from the femurs, 109 from the tibias and 25 from the fibulas All the samples were tested wet The actual values for strength were not stated the results of the tests being graphically represented by histograms

The average ultimate tensile strength in all three long bones of the lower limb was greatest in the specimens from the middle third of the shaft The lowest strength among the femoral specimens was in the proximal and distal thirds which were approximately equal The specimens from the proximal third of the tibia and the distal third of the fibula had the lowest ultimate strength With respect to the entire bone the tibial specimens exhibited the highest and the femoral specimens the lowest average strength The average elongation or strain was greatest in the proximal third of the femur and fibula and the middle third of the tibia The least elongation occurred in samples from the distal third of the femur and fibula and in the proximal third of the femur Considered as a whole the fibular samples showed the greatest and the femoral samples the least average elongation or strain

One of the most interesting findings was the high tensile strength exhibited by the fibula which per unit area, exceeded that of the femur Another important point was the demonstration that the strength of the shaft of the three long bones of the lower limb showed considerable variation This is a factor that might have clinical importance with respect to the selection of bones from which to take specimens for grafting

The ultimate tensile strength and other physical properties

of human compact bone have also been investigated by Dempster and Liddicoat (1952) Their study was based upon samples obtained from cleaned and dried adult macerated preparations Samples from 13 femurs ten tibias, and five humeri were tested No data were available on the age sex, race or method of preservation of the bodies from which the bones were taken None of the bones showed obvious pathology

Two types of tension samples, spindles and rectangular prisms, five to six inches long approximately ¼ of an inch thick and with a two inch reduced area having a cross section of 0 02 to 0 03 in² were used The samples were taken from the middle third of the bones generally the anterior aspect Test samples were considered as dry or as wet, after soaking for 24 hours in water The only fresh samples were from a middle aged male which had been refrigerated for some days The samples were tested under direct tension in the long axis of the bone Several types of testing machines all calibrated to an accuracy of ± 1% were used Generally the loading speed was about 400 lbs/min

The strain occurring in the samples was measured with Porter-Lipp Huggenberger or Tuckermann extensometers fixed along a standard gage length in the reduced area of the samples Two extensometers one on each side of the sample compensated for possible eccentricities in loading and gave more consistent records than those obtained from a single extensometer How ever both one and two extensometer records were obtained

Dempster and Liddicoat found that the ultimate tensile strength and stress of the 14 wet specimens was 11428 ± 1540 lbs/in² The corresponding value for the 11 dry specimens was 17090 ± 3940 lbs/in One of their figures summarizing the data of previous investigators shows that the ultimate tensile strength and stress of wet bone is 3000 to 5000 lbs/in less than that of dry bone

Data obtained from the extensometers applied to the specimens during the tests were used in determining the modulus of elasticity of the specimens and for plotting stress strain curves The stress strain for wet and dry compact bone under tension (Figure 40a) was similar to those previously illustrated by Evans and Lebow (Figure 38)

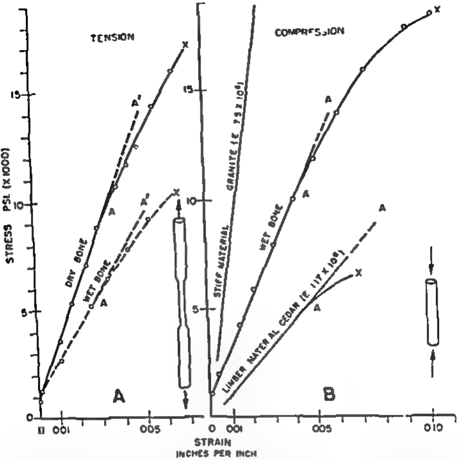

Figure 40 Stress strain curves for human compact bone (From the original Figure 1 Dempster and Liddicoat *Am J Anat* *91* 336 1952) A = proportional limit X = breaking point O A = slope of perfectly elastic range A X = range of plastic deformation

The stress strain relationships of bony material under tension have been investigated by several individuals Wertheim (*loc cit*) concluded that bone followed the law of proportionality to weight as do inorganic bodies and wood Thus plotting weight (stress) against elongation (strain) gave a straight line Soft tissues in their natural state of humidity did not behave this way their stress strain curve approaching a hyperbole The direct relation of stress and strain was especially marked when the bone had previously been well dried In fresh bone the co efficient of elasticity increased little with load and the elongations did not increase strictly with load but showed a less directly pro

of human compact bone have also been investigated by Dempster and Liddicoat (1952) Their study was based upon samples obtained from cleaned and dried adult macerated preparations Samples from 13 femurs ten tibias, and five humeri were tested No data were available on the age, sex race or method of preservation of the bodies from which the bones were taken None of the bones showed obvious pathology

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TABLE VI

AVERAGE ULTIMATE TENSILE STRENGTH (LBS/IN^2) OF HUMAN COMPACT BONE LOADED IN THE DIRECTION OF THE LONG AXIS

Bone	*Author*	*Sex*	*No of Samples*	*Condition of Sample*	*Ultimate Tensile Strength*
Humerus	Rauber 1876	Male	11	Fresh	14560 (8880–18650)
	Hulsen 1896	Male	4	Fresh	15000 (13380–16280)
		Male	4	Dry unembalmed	15550 (11850–17480)
Femur	Wertheim 1847	Male	2	Fresh	12650 (10380–14930)
		Female	2	Fresh	9135 (9100–9770)
	Rauber 1876	Male	19	Fresh	13850 (9240–18840)
		Female	[illegible]	Fresh	9020 (8170–9950)
		Male	6	Dry unembalmed	16460 (14220–18670)
	Hulsen 1896	Male	6	Dry unembalmed	16850 (14260–21720)
	Carothers Smith & Calabrisi 1949	?	11	Dry embalmed	22000 (16600–31500)
	Evans & Lebow 1951	Male	121	Wet embalmed	11840 (6990–15500)
		Male	121	Dry embalmed	15330 (8740–21510)
	Evans (unpublished)	Female	58	Wet embalmed	11250 (3640–18130)
Tibia	Rauber 1876	Male	8	Fresh	17640 (15110–24170)
	Hulsen 1896	Male	4	Fresh	15040 (12880–16250)
	Evans (unpublished)	Male	75	Wet unembalmed	13159 (5199–21300)
Fibula	Wertheim 1847	Male	2	Fresh	13765 (6160–21370)
		Female	2	Fresh	9640 (4692–14593)
	Evans (unpublished)	Male	15	Wet unembalmed	14640 (8825–19890)
Not stated	Dempster & Liddicoat 1952	?	11	Dry	17090 ± 3940
		?	14	Wet	11428 ± 1540

The average ultimate tensile strength regardless of source obtained by various investigators (Table VI) varied from 9020 - 15040 lbs/in^2 for fresh bone, 15550 - 16850 lbs/in^2 for dry unembalmed bone, 11250 - 11840 lbs/in^2 for wet embalmed bone and 15330 - 22000 lbs/in^2 for dry embalmed bone. Dempster and Liddicoat (*loc. cit.*) reported 11070 ± 3940 and 11428 ± 1540 lbs/in^2 for dry and wet bone respectively. However they did not state whether or not the bone was embalmed. All investiga-

portional relation Wertheim accounted for this difference on the basis of the moisture in the fresh bone

Wertheim s conclusions are verified by the stress strain curves illustrated by Evans and Lebow (Figure 38) and Dempster and Liddicoat (Figure 40) Both teams of investigators found that the stress strain curve for dry bone in tension is essentially a straight line to failure but that for wet bone deviates from a straight line as the rupture point is approached In comparing statements made by different investigators regarding the behavior of bony tissue under tension one must know whether the specimen was tested to the elastic limit, the proportional limit or the ultimate limit each of which is a point on the stress strain curve The elastic limit is the point up to which the specimen will return to its original dimensions when the load is removed The proportional limit indicates the point up to which strain or deformation is directly proportional to load, i e , the straight line portion of the stress strain curve The ultimate limit is the point at which the maximum stress is reached The exact place of each of these points on the stress strain curve varies with the material being analyzed and in some instances two points may coincide

Evans and Lebow (1951) for example found that the tensile stress strain curve for dry bone (Figure 38) is a straight line to failure which means that the proportional limit and the ultimate limit were the same However the stress strain curve for bone under tension illustrated by Dempster and Liddicoat (Figure 40a) shows a slight deviation from a straight line before failure occurs The point (A) at which the curve starts to deviate from a straight line marks the proportional limit of the sample Up to this point strain (elongation) was proportional to stress (load) but beyond this point strain increased more rapidly than stress In addition the strain or deformation beyond the proportional limit was permanent i e the bone did not return to its original dimensions after removal of the load The load or stress at which the proportional limit was reached was about half the breaking load or stress Dempster and Liddicoat also found that the strain or deformation produced by a given stress (load) was greater in wet than in dry bone

TABLE VI

AVERAGE ULTIMATE TENSILE STRENGTH (LBS/IN^2) OF HUMAN COMPACT BONE LOADED IN THE DIRECTION OF THE LONG AXIS

Bone	*Author*	*Sex*	*No. of Samples*	*Condition of Sample*	*Ultimate Tensile Strength*
Humerus	Rauber 1876	Male	11	Fresh	14560 (8850–18650)
	Hulsen 1896	Male	4	Fresh	15000 (13380–16280)
		Male	4	Dry unembalmed	15550 (11850–17480)
Femur	Wertheim 1847	Male	2	Fresh	12650 (10380–14930)
		Female	2	Fresh	9135 (9100–9770)
	Rauber 1876	Male	19	Fresh	13850 (9210–18840)
		Female	5	Fresh	9020 (8170–9950)
		Male	6	Dry unembalmed	16160 (14220–18670)
	Hulsen 1896	Male	6	*Dry unembalmed*	*16850 (14260–21720)*
	Carothers, Smith & Calabrisi 1949	?	11	Dry embalmed	22000 (16600–31500)
	Evans & Lebow 1951	Male	121	Wet embalmed	11840 (6930–15500)
		Male	121	Dry embalmed	15330 (8740–21510)
	Evans (unpublished)	Female	58	Wet embalmed	11250 (3640–18130)
Tibia	Rauber 1876	Male	8	Fresh	17640 (15110–21170)
	Hulsen 1896	Male	4	Fresh	15040 (12880–16250)
	Evans (unpublished)	Male	75	Wet unembalmed	13150 (5190–21300)
Fibula	Wertheim 1847	Male	2	Fresh	13765 (6160–21370)
		Female	2	Fresh	9640 (4692–14593)
	Evans (unpublished)	Male	15	Wet unembalmed	14640 (8825–19890)
Not stated	Dempster & Liddicoat 1952	?	11	Dry	17090 ± 3940
		?	14	Wet	11428 ± 1540

The average ultimate tensile strength, regardless of source, obtained by various investigators (Table VI) varied from 9020 - 15040 lbs/in² for fresh bone, 15550 – 16850 lbs/in² for dry unembalmed bone, 11250 – 11840 lbs/in² for wet embalmed bone, and 15330 – 22000 lbs/in² for dry embalmed bone. Dempster and Liddicoat (*loc. cit.*) reported 11070 ± 3940 and 11428 ± 1540 lbs/in² for dry and wet bone respectively. However, they did not state whether or not the bone was embalmed. All investiga-

tors who studied the influence of moisture found that drying increased the ultimate tensile strength of bone

Another factor to be considered in comparing the results various authors give for ultimate tensile strength is whether or not their samples were from unembalmed or embalmed bone Calibrisi and Smith (1951) reported that embalming reduces the compressive strength of fresh bone by 13%, but the author knows of no similar study on the influence of embalming on the tensile strength of bone However the average ultimate tensile strength of wet embalmed bone as reported by different authors (Table VI) is usually considerably less than that for fresh bone

Rauber (*loc cit*) found a decrease in the average tensile strength of bone with the increasing age of the individual from whom the samples were obtained but Wertheim (*loc cit*) and Evans and Lebow (1951) reported no consistent relation between the age of the individual and strength of bone The latter two authors also found no significant difference between the average ultimate tensile strength of right and left femurs but most authors did not designate the side of the body from which their samples were obtained Among the long bones of the lower limb the tibia is the strongest per unit area the fibula is intermediate, and the femur is weakest

Relatively few samples of female bone have been tested but they are generally weaker per unit area than those of males From the figures given by Wertheim (*loc cit*) and by Rauber (*loc cit*) the author computed that fresh compact bone from male femurs has an average ultimate tensile strength 34% and 54% respectively greater than that from female femurs However, the average tensile strength of samples from wet embalmed male femurs (data from Evans and Lebow, 1951) is only 5% greater than similar samples from female femurs (data from Evans unpublished)

Wertheim (*loc cit*) made some measurements of the strain or elongation occurring in his samples under direct tension but the strain was not determined at the fracture load Consequently, his results cannot be compared with those reported by Evans and Lebow (1951)

Comparison of the average ultimate tensile strength of hu

TABLE VII

AVERAGE ULTIMATE TENSILE STRENGTH (LBS/IN²) OF NONHUMAN COMPACT BONE

Bone	Author	Animal	No. of Samples	Condition of Samples	Direction of Loading	Ultimate Tensile Strength
Femur	Hulsen 1896	Ox	4	Fresh	Long axis of bone	1[illegible]130 (14220-1[illegible]920)
		Calf	3	Fresh	Long axis of bone	11180 (10[illegible]00-12210)
		Wolf	4	Fresh	Long axis of bone	1[illegible]300 (13020-17[illegible]0)
Tibia	Rauber 1876	Ox	7	Fresh	Long axis of bone	16200 (14220-21330)
		Ox	6	Dry	Long axis of bone	17670 ([illegible]070-21000)
		Calf	5	Fresh	Long axis of bone	12600 (8[illegible]30-17[illegible]0)
		Domestic pig	1	Fresh	Long axis of bone	10380
		Wild pig	1	Fresh	Long axis of bone	14630
	Hulsen 1896	Ox	4	Fresh	Long axis of bone	16770 (1[illegible]670-180[illegible]0)
		Ox	5	Fresh	Transverse axis of bone	12960 (11010-1[illegible]110)
		Ox	3	Decalcified	Long axis of bone	2670 (2[illegible]80-2810)
		Ox	3	Decalcified	Transverse axis of bone	1[illegible]30 (920-2100)

TABLE VIII

AVERAGE ULTIMATE COMPRESSIVE STRENGTH (LBS/IN²) OF HUMAN COMPACT BONE

Bone	*Author*	*Sex*	*No of Samples*	*Condition of Sample*	*Sample Form*	*Direction of Load*	*Compressive Strength*	
Humerus	Rauber 1876	M	13	Fresh	Cube	Long axis of bone	19820	(16430–27558)
		M	5	Dry	Cube	Long axis of bone	19410	(18650–20970)
		M	6	Dry	Cube	Perpendicular to long axis of bone	16060	(14220–18050)
	Hulsen 1896	M	2	Fresh	Cube	Long axis of bone	29030	(28890–29160)
		M	2	Fresh	Cube	Perpendicular to long axis of bone	23460	(20170–26740)
		M	6	Fresh	Ring	Long axis of bone	29020	(22110–36160)
	Dempster & Liddicoat 1952	?	3	Dry	Cube	Long axis of bone	28277	(26583–31200)
		?	3	Dry	Cube	Radial axis of bone	16916	(11850–21633)
		?	3	Dry	Cube	Tangential axis of bone	16550	(12183–18900)
Femur	Rauber 1876	M	17	Fresh	Cube	Long axis of bone	20430	(16160–24880)
		M	12	Dry	Cube	Long axis of bone	26990	(21330–32990)
		M	6	Dry	Cube	Perpendicular to long axis of bone	25310	(23660–26160)
		F	5	Fresh	Cube	Long axis of bone	18580	(17080–20530)
	Hulsen 1896	M	5	Dry	Ring	Long axis of bone	29910	(21050–36800)
	Carothers Smith & Calabrisi 1949	?	1	Dry	Piece of whole shaft (Type A)	Perpendicular to direction of fibers	28100	
		?	3	Dry	Hollow cylinder (Type B)	Perpendicular to direction of fibers	24200	(20800–28600)
		?	1	Dry	Block	Perpendicular to direction of fibers	23000	
		?	8	Embalmed	Hollow cylinder (Type B)	Direction of fibers	25350	(20900–31700)

TABLE VIII (cont'd)

AVERAGE ULTIMATE COMPRESSIVE STRENGTH (LBS/IN²) OF HUMAN COMPACT BONE

Bone	*Author*	*Sex*	*No. of Samples*	*Condition of Sample*	*Sample Form*	*Direction of Load*	*Compressive Strength*	
Femur	Calabrisi & Smith 1951	M	[illegible]	Fresh	Hollow cylinder (Type B)	Direction of fibers	301[illegible]0	(29300–31000)
		M	2	Embalmed		Direction of fibers	2[illegible]300	(23000–2[illegible]600)
		F	1	Fresh		Direction of fibers	31000	
		F	1	Embalmed		Direction of fibers	23[illegible]00	
	Dempster & Liddicoat 1952	?	4	Dry	Cube	Long axis of bone	30[illegible]	(27317–331[illegible]0)
		?	4	Dry	Cube	Radial axis of bone	21[illegible]77	(17[illegible]3–27[illegible]07)
		?	4	Dry	Cube	Tangential axis of bone	20266	(1[illegible]200–21330)
Tibia	Rauber 1876	M	7	Fresh	Cube	Long axis of bone	23880	(22220–2[illegible]240)
		M	5	Dry	Cube	Long axis of bone	24[illegible]10	(23660–26160)
		M	5	Dry	Cube	Perpendicular to long axis of bone	17910	(16[illegible]00–19[illegible]00)
	Hulsen 1896	M	4	Fresh	Cube	Long axis of bone	29760	(26600–32130)
	Carothers, Smith & Calabrisi 1949	?	4	Embalmed	Hollow cylinder (Type B)	Direction of fibers	2777[illegible]	(26600–29200)
		?	1	Embalmed	Piece of whole shaft (Type A)	Direction of fibers	20100	
	Calabrisi & Smith 1951	M	2	Fresh	Hollow cylinder (Type B)	Direction of fibers	22100	(21100–23100)
		M	[illegible]	Embalmed	"	Direction of fibers	231[illegible]0	(20200–26700)
		F	4	Fresh	"	Direction of fibers	26870	(23300–31100)
		F	2	Embalmed		Direction of fibers	22010	(17020–27000)
		F	7	Fresh	Solid miniature cylinder	Direction of fibers	2[illegible]200	(1[illegible]00–31300)
Fibula	Carothers, Smith & Calabrisi 1949	?	1	Dry	Hollow cylinder (Type B)	Perpendicular to direction of fibers	23200	

TABLE VIII (cont'd)

AVERAGE ULTIMATE COMPRESSIVE STRENGTH (LBS/IN²) OF HUMAN COMPACT BONE

Bone	*Author*	*Sex*	*No. of Samples*	*Condition of Sample*	*Sample Form*	*Direction of Load*	*Compressive Strength*	
Femur	Calabrisi & Smith 1951	M	2	Fresh	Hollow cylinder (Type B)	Direction of fibers	30150	(29300–31000)
		M	2	Embalmed		Direction of fibers	24300	(23000–25600)
		F	1	Fresh		Direction of fibers	31000	
		F	1	Embalmed		Direction of fibers	23500	
	Dempster & Liddicoat 1952	?	4	Dry	Cube	Long axis of bone	30580	(27317–33150)
		?	4	Dry	Cube	Radial axis of bone	21577	(17183–27607)
		?	4	Dry	Cube	Tangential axis of bone	20260	(15900–21330)
Tibia	Rauber 1876	M	7	Fresh	Cube	Long axis of bone	23880	(22220–25210)
		M	5	Dry	Cube	Long axis of bone	21510	(23040–26140)
		M	5	Dry	Cube	Perpendicular to long axis of bone	17010	(11100–19000)
	Hulsen 1896	M	1	Fresh	Cube	Long axis of bone	29700	(26600–32150)
	Carothers, Smith & Calabrisi 1949	?	4	Embalmed	Hollow cylinder (Type B)	Direction of fibers	27775	(26400–29900)
		?	1	Embalmed	Piece of whole shaft (Type A)	Direction of fibers	20100	
	Calabrisi & Smith 1951	M	2	Fresh	Hollow cylinder (Type B)	Direction of fibers	22100	(21100–23100)
		M	2	Embalmed		Direction of fibers	23150	(20200–26700)
		F	4	Fresh		Direction of fibers	28570	(25300–31100)
		F	2	Embalmed		Direction of fibers	22010	(17090–27000)
		F	7	Fresh	Solid miniature cylinder	Direction of fibers	2[illegible]300	(18600–31300)
Fibula	Carothers, Smith & Calabrisi 1949	?	1	Dry	Hollow cylinder (Type B)	Perpendicular to direction of fibers	21900	

TABLE VIII

AVERAGE ULTIMATE COMPRESSIVE STRENGTH (LBS/IN²) OF HUMAN COMPACT BONE

Bone	*Author*	*Sex*	*No. of Samples*	*Condition of Sample*	*Sample Form*	*Direction of Load*	*Compressive Strength*	
Humerus	Rauber 1876	M	13	Fresh	Cube	Long. axis of bone	19820	(16430–27558)
		M	5	Dry	Cube	Long. axis of bone	19110	(18650–20970)
		M	6	Dry	Cube	Perpendicular to long. axis of bone	16060	(11220–18050)
	Hulsen 1896	M	2	Fresh	Cube	Long. axis of bone	29030	(28890–29100)
		M	2	Fresh	Cube	Perpendicular to long. axis of bone	23460	(20170–26710)
		M	6	Fresh	Ring	Long. axis of bone	29020	(22110–36160)
	Dempster & Liddicoat 1952	?	3	Dry	Cube	Long. axis of bone	28277	(26583–31200)
		?	3	Dry	Cube	Radial axis of bone	16916	(11850–21633)
		?	3	Dry	Cube	Tangential axis of bone	16550	(12183–18900)
Femur	Rauber 1876	M	17	Fresh	Cube	Long axis of bone	20130	(16160–24880)
		M	12	Dry	Cube	Long. axis of bone	26990	(21330–32990)
		M	6	Dry	Cube	Perpendicular to long. axis of bone	25310	(23600–26160)
		F	5	Fresh	Cube	Long. axis of bone	18580	(17680–20030)
	Hulsen 1896	M	5	Dry	Ring	Long. axis of bone	29910	(21050–36800)
	Carothers, Smith & Calabrisi 1949	?	1	Dry	Piece of whole shaft (Type A)	Perpendicular to direction of fibers	28100	
		?	3	Dry	Hollow cylinder (Type B)	Perpendicular to direction of fibers	24200	(20800–28600)
		?	1	Dry	Block	Perpendicular to direction of fibers	23000	
		?	8	Embalmed	Hollow cylinder (Type B)	Direction of fibers	25350	(20900–31700)

TABLE IX

AVERAGE ULTIMATE COMPRESSIVE STRENGTH (LBS/IN²) IN NONHUMAN BONE

Bone	*Author*	*Animal*	*No. of Samples*	*Condition of Samples*	*Type of Sample*	*Direction of Loading*	*Ultimate Compressive Strength*	
Femur	Hulsen 1896	Ox	5	Fresh	Ring	Long axis of bone	29070	(26000–33310)
		Calf	3	Fresh	Ring	Long axis of bone	17610	(1[illegible]–18180)
		Domestic pig	2	Fresh	Ring	Long axis of bone	17630	(14250–21010)
		Wolf	3	Fresh	Ring	Long axis of bone	25370	([illegible]–28400)
Tibia	Rauber 1876	Ox	2	Fresh	Cube	Long axis of bone	18920	(17310–1[illegible])
		Ox	6	Dry	Cube	Perpendicular to long axis of bone	16060	(11250–18170)
		Calf	4	Fresh	Cube	Long axis of bone	17130	(1[illegible]40–19170)
		Calf	5	Dry	Cube	Long axis of bone	21580	(18310–2[illegible]240)
		Domestic pig	1	Fresh	Cube	Long axis of bone	16080	
		Wild pig	1	Fresh	Cube	Long axis of bone	20110	
Fibula	Hulsen 1896	Ox	9	Fresh	Cube	Perpendicular to long axis of bone	28180	(20970–33170)

man compact bone (Table VI) with that of similar samples from the leg bones of other mammals (Table VII) is instructive. Under a similar type of loading the samples from the human femur were weaker than those from the femur of an ox and a wolf but stronger than those from a calf femur. Rauber found that the fresh samples from the human tibia were stronger than those he tested from the tibia of an ox, a calf and a domestic and a wild pig, but Hulsen reported that his samples from the ox tibia were the stronger. Humeral and fibular samples from nonhuman mammals were not tested. Examination of the tables also reveals that although the average ultimate tensile strength of the human samples may be less than those of other mammals their range of variation overlaps that of the nonhuman mammals studied.

As far as the author is aware the tensile strength of spongy bone has not been determined for any animal. The probable reason is the difficulty of preparing samples that can be held in a testing machine.

Compressive Stress and Strain of Compact Bone

The ultimate compressive strength of compact bone of man and other mammals (Tables VIII and IX) has been investigated by the same authors, except for Wertheim (*loc. cit.*) and Evans and Lebow (*loc. cit.*) who studied its tensile strength. However the results obtained by the various workers are not as directly comparable with one another as those for tensile strength because of differences in the shape of the test samples and the direction of loading with respect to the various axes of the bone. The compressive strength of bone is also influenced by drying and embalming.

Rauber (1876) found that the average ultimate compressive strength of fresh cubes loaded in the direction of the long axis of the bone was greatest (23880 lbs/in.) in samples from the tibia, intermediate (22230 lbs/in.) in femoral samples and least (19820 lbs/in.) in humeral samples. Drying cubes of unembalmed humeral bone slightly reduced the compressive strength under similar loading to 19410 lbs/in. but increased the compressive strength (26990 lbs/in.^{2}) of the femoral samples. The

to 0.000002 in/in. The accuracy of the wire and optical lever strain gages was 3% and 0.5% respectively.

The results obtained for compressive strength depended on the direction of loading of the samples. The ultimate compressive strength for the single type A air dried femoral sample loaded perpendicular to the direction of the fibers was 28100 lbs/in^2. Under similar loading the average compressive strength of three air dried type B femoral samples was 24200 lbs/in^2 (20800 - 28600 lbs/in^2) while a single air dried block from a femur had a strength of 23000 lbs/in^2. The ultimate compressive strength of a single air dried type B fibular sample subjected to similar loading was 23200 lbs/in^2.

The strength of samples loaded in the direction of their fibers was generally greater. Thus, eight embalmed but not air dried type B femoral samples had an average ultimate compressive strength of 25350 lbs/in^2 while four similar tibial samples had an average strength of 27775 lbs/in^2. The ultimate compressive strength of a single embalmed but not air dried type A tibial sample was 20400 lbs/in^2.

The influence of embalming upon the compressive strength of compact bone was investigated by Calabrisi and Smith (1951) in type B samples from the middle third of amputated femurs and tibias. Two samples, as nearly identical as possible, were machined from each of seven bones, one member of each pair being tested in the fresh condition while the other was put in embalming fluid for 43 to 50 days before testing. Several miniature cylindrical samples about 1/8 inch in diameter and 3/8 inch long as well as two tubular (Type B) samples were also tested.

It was found that the average ultimate compressive strength of the regular sized fresh and embalmed samples was 27200 lbs/in^2 and 23300 lbs/in^2, respectively. The miniature samples had an average compressive strength of 25200 lbs/in^2 (18900 - 31300 lbs/in^2) just 4.8% less than that of the two tubular samples (26000 lbs/in^2).

Calabrisi and Smith therefore concluded that embalming reduces the compressive strength of compact bone about 13% and that miniature samples are suitable for testing the compressive strength of bone. They believed that the variation in compressive

compressive strength (17910 lbs/in²) of dry tibial cubes loaded perpendicular to the long axis of the bone was less than that of similar fresh cubes loaded parallel with their long axis

The cubes of compact bone tested by Hulsen (1896) were generally stronger than Rauber's samples Thus the average compressive strength of fresh humeral cubes, loaded in the direction of the long axis of the bone was 29030 lbs/in², while similarly loaded tibial cubes had a strength of 29760 lbs/in² Fresh humeral cubes loaded perpendicular to the long axis of the bone had an average ultimate strength of 23460 lbs/in² Fresh humeral rings loaded parallel with the long axis of the bone had an average compressive strength of 29020 lbs/in² while similarly loaded fresh tibial rings were slightly stronger (29760 lbs/in²) The average strength of similarly loaded dry femoral rings was about the same (29910 lbs/in²)

Carothers Smith, and Calabrisi (*loc cit*) made 23 compression tests of samples of compact bone from the shaft of 14 femurs eight tibias and one fibula Eight of the samples were from femurs which had been air dried for several years while the remaining samples were taken from the bones and tested within a year after embalming of the body The bones were the same as those from which the tensile samples had been taken

Three types of samples were tested Type A samples were pieces cut from the middle third of the shaft with their ends machined flat and parallel to within 0 001 inch Because they were sections from the entire shaft they were cylinders with walls of varying thickness Type B samples were hollow cylinders taken from the middle third of the shaft but machined to have walls of uniform thickness The dimensions within a sample were uniform to within 0 001 inch The cylinders consisted of smooth compact bone with their long axis approximately parallel with that of the marrow cavity of the intact bone Type C samples were checker like leaves of compact bone

The samples were tested in hydraulic testing machines calibrated to an accuracy of ± 1% The deformation or strain occurring in a sample during a test was measured with electrical wire resistance or optical lever strain gages capable of measuring strain

to 0.000002 in/in. The accuracies of the wire and optical lever strain gages was 3% and 0.5% respectively.

The results obtained for compressive strength depended on the direction of loading of the samples. The ultimate compressive strength for the single type A air dried femoral sample loaded perpendicular to the direction of the fibers was 25100 lbs/in^2. Under similar loading the average compressive strength of three air dried type B femoral samples was 24200 lbs/in^2 (20800 - 28600 lbs/in^2) while a single air dried block from a femur had a strength of 23000 lbs/in^2. The ultimate compressive strength of a single air dried type B fibular sample subjected to similar loading was 23200 lbs/in^2.

The strength of samples loaded in the direction of their fibers was generally greater. Thus eight embalmed but not air dried type B femoral samples had an average ultimate compressive strength of 25350 lbs/in^2 while four similar tibial samples had an average strength of 27775 lbs/in. The ultimate compressive strength of a single embalmed but not air dried type A tibial sample was 20400 lbs/in.

The influence of embalming upon the compressive strength of compact bone was investigated by Calabrisi and Smith (1951) in type B samples from the middle third of amputated femurs and tibias. Two samples as nearly identical as possible were machined from each of seven bones, one member of each pair being tested in the fresh condition while the other was put in embalming fluid for 43 to 50 days before testing. Several miniature cylindrical samples about $\frac{1}{8}$ inch in diameter and $\frac{3}{8}$ inch long as well as two tubular (Type B) samples were also tested.

It was found that the average ultimate compressive strength of the regular sized fresh and embalmed samples was 27200 lbs/in. and 23300 lbs/in., respectively. The miniature samples had an average compressive strength of 25200 lbs/in. (18900 - 31300 lbs/in.) just 4.8% less than that of the two tubular samples (26000 lbs/in.).

Calabrisi and Smith therefore concluded that embalming reduces the compressive strength of compact bone about 13% and that miniature samples are suitable for testing the compressive strength of bone. They believed that the variation in compressive

strength, because of embalming was similar to the variation from bone to bone, as evidenced by the fact that the average compressive strength of the embalmed samples tested in an earlier study (Carothers, Smith and Calabrisi) was about 10% greater than the embalmed samples tested in the second study (Calabrisi and Smith).

The effect of direction of loading on the results obtained for compressive strength was clearly demonstrated by Dempster and Liddicoat (*loc. cit.*) who tested cubes of compact bone by loading them in the direction of their long, their radial, and their tangential axis. All samples were from dry embalmed human humeri and femora. The ultimate compressive strength was least (20266 lbs/in^2) in the tangentially loaded femoral cubes and greatest (30586 lbs/in^2) in the longitudinally loaded ones. The same was true for the humeral cubes, the actual values for the tangentially and longitudinally loaded samples being 16550 lbs/in^2 and 28277 lbs/in^2 respectively.

All investigators who have compared the ultimate compressive strength of wet and dry samples of compact bone found that the latter were the stronger. Compact bone also has a greater compressive strength when loaded parallel with the long axis of the bone than when loaded perpendicular to the long axis. Dempster and Liddicoat found no significant difference between the radial and tangential compressive strength. Embalming reduces the compressive strength of compact bone about 13%.

The ultimate compressive strength of compact bone is considerably greater than its tensile strength and, according to Rauber (*loc. cit.*), *decreases less with aging*. The same author found that both tensile and compressive strength of compact bone are decreased by heating. The average ultimate compressive strength and range of variation of compact human bone is similar to that of other mammals. Carothers, Smith and Calabrisi (1949) found that embalming increased the strain of intact femurs of young male albino rats.

Dempster and Liddicoat (*loc. cit.*) are, as far as the author is aware, the only investigators who have published the stress-strain curve for wet compact bone under compression (Figure 40b). The curve was a straight line as far as the proportional

limit, beyond which it resembled the tensile stress strain curve (Figures 38 40 a) in deviating from a straight line as the failure point was approached Comparison of the stress strain curves for wet and dry compact bone under tension and compression (Figure 40) shows that drying increases the tensile strength of bone and that bone is considerably stronger in compression than in tension The ultimate tensile strength of the wet bone was slightly over 10 000 lbs/in² while the compressive strength of corresponding bone was approximately 18 000 lbs/in² The ultimate strain for wet bone was approximately 0 007 in/inch in tension and 0 11 in/inch in compression

Compressive Strength of Spongy Bone

Relatively few studies have been made of the compressive strength of spongy bone Rauber (*loc cit*) reported that the average compressive strength, tested parallel with the long axis of the bone of four fresh cubes of spongy bone from the lumbar vertebrae of adult men was 1190 lbs/in² (920 1350) Four similarly tested fresh cubes from the femoral condyles of adult men had an average strength of 1360 lbs/in (1130 1700)

Hardinge (1949) investigated the compressive strength of spongy bone from the head and neck of 94 femurs from embalmed dissecting room cadavers Five transverse sections approximately ¼ inch thick were cut from each femur and the strength of the spongy bone determined by measuring in pounds the force required to crush the bone The crushing was done with a flat nosed metal punch ¼ inch in diameter and driven by a constant speed motor The punch had an accuracy of 0 2% and was adjusted to penetrate the bone to a depth of ⅛ inch The first section of the bone was through the head and the fifth through the junction of the head and neck The test specimens were kept in water to maintain a constant moisture content One specimen from section one 13 from sections two three and four and five from section five of each femur were tested Measurements were made on specimens from 31 right and 29 left femurs of males and from 16 right and 18 left femurs of females The average age of the males was 60 9 years and the females 70 8 years

strength because of embalming was similar to the variation from bone to bone as evidenced by the fact that the average com pressive strength of the embalmed samples tested in an earlier study (Carothers Smith and Calabrisi) was about 10% greater than the embalmed samples tested in the second study (Cala brisi and Smith)

The effect of direction of loading on the results obtained for compressive strength was clearly demonstrated by Dempster and Liddicoat (*loc cit*) who tested cubes of compact bone by load ing them in the direction of their long their radial, and their tangential axis All samples were from dry embalmed human humeri and femora The ultimate compressive strength was least (20266 lbs/in^2) in the tangentially loaded femoral cubes and greatest (30586 lbs/in^2) in the longitudinally loaded ones The same was true for the humeral cubes the actual values for the tan gentially and longitudinally loaded samples being 16550 lbs/in and 28277 lbs/in^2 respectively

All investigators who have compared the ultimate compres sive strength of wet and dry samples of compact bone found that the latter were the stronger Compact bone also has a greater compressive strength when loaded parallel with the long axis of the bone than when loaded perpendicular to the long axis Dempster and Liddicoat found no significant difference between the radial and tangential compressive strength Embalming re- duces the compressive strength of compact bone about 13%

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Dempster and Liddicoat (*loc cit*) are as far as the author is aware the only investigators who have published the stress strain curve for wet compact bone under compression (Figure 40b) The curve was a straight line as far as the proportional

show that it is a great deal less than that of compact bone. This should not be surprising because the load supporting part of an intact long bone or vertebral body is the compact rather than the spongy bone. This proposes the question of the mechanical function, if any, of spongy bone, particularly with reference to the proximal end of the femur upon which the trajectorial theory of the organization of the spongy bone is primarily based.

In previous discussions of the trajectorial theory of bone architecture it was pointed out that various trabeculae in the neck of the femur have been considered to be tensile or compressive resistant. However, in all probability spongy or trabecular bone is not a force or load transmitting portion of the bone at all. Rather it would seem, as in the cross piece of an I beam, that the function of spongy bone may be to separate or hold apart the actual load carrying parts (compacta) of the intact bone. In fractures of the femoral neck and trochanteric region experimentally produced by loading the greater trochanter it was frequently found that the trochanter itself was undamaged.

Anatomically the greater trochanter consists of a thin shell of compact bone filled with spongy bone and the fact that it can support a load, sometimes more than 1000 lbs., sufficiently great to fracture the bone without the trochanter itself being damaged is of considerable interest. In x-rays of clinical fractures of the femoral neck and trochanteric region, often arising from falls on the trochanter, the trochanter frequently appears to be undamaged. In the living body the soft tissue overlying the greater trochanter would absorb much of the energy of a fall but such was not the case in the femoral fractures produced in dry bones by static or dynamic loading of the greater trochanter. In these cases the energy was very likely absorbed by the spongy bone itself, which is a good energy absorbing material.

Shearing Strength of Bone

The direct shearing strength of bone was first tested by Rauber (*loc. cit.*) who designed a special tool which subjected standardized specimens of compact and spongy bone to a double shearing action. Rauber emphasized that in shearing tests bending can only occur in two directions but if tested in the third

Contrary to his belief Hardinge did not determine the strength of cancellous bone from the femur but only the load or force necessary to crush it. He also makes the astonishing statement that "the force required to crush the bone without producing compression was measured in pounds." This statement is obviously an error because it is physically impossible to produce crushing in any material without compression.

Hardinge recorded the average force (in pounds) needed to crush various regions in each of the five cross sections of the femoral head and neck. From this data the author has calculated the actual compressive strength or stress (lbs/in²) of the spongy bone in the different sections. In doing this the cross section area of the punch was determined from the formula for the area of a circle and it was assumed that the force delivered by the punch was uniformly distributed over its cross section area.

The author's computation of the average ultimate compressive strength of the spongy bone in the different sections was 3380 lbs/in² for section one, 4317 lbs/in² (2700-6120) for section two, 4383 lbs/in² (2100-7240) for section three, 3588 lbs/in² (2480-5380) for section four, and 3328 lbs/in² (2620-5160) for section five.

The above data show that the average compressive strength was greatest in section three although section two was only very slightly less. The samples from section three also exhibited the greatest range of variation. Section five was the weakest. The average strength of section three was 31% greater than that of section five.

Hardinge found a band of greatest compressive strength extending vertically from the most superior point on the head, i.e., the weight supporting area in the erect posture, to the junction of the head and the inferior aspect of the neck. This band corresponded to the so-called compression resisting lamellae of the trajectorial theory of bone architecture. Little difference was found in the compressive strength of specimens from right and left femurs. The data were not presented in such a way that the influence of sex on the strength of bones can be computed.

All of the studies of the compressive strength of spongy bone

show that it is a great deal less than that of compact bone. This should not be surprising because the load supporting part of an intact long bone or vertebral body is the compact rather than the spongy bone. This proposes the question of the mechanical function, if any, of spongy bone, particularly with reference to the proximal end of the femur upon which the trajectorial theory of the organization of the spongy bone is primarily based.

In previous discussions of the trajectorial theory of bone architecture it was pointed out that various trabeculae in the neck of the femur have been considered to be tensile or compressive resistant. However, in all probability spongy or trabecular bone is not a force or load transmitting portion of the bone at all. Rather it would seem, as in the cross piece of an I beam, that the function of spongy bone may be to separate or hold apart the actual load carrying parts (compacta) of the intact bone. In fractures of the femoral neck and trochanteric region experimentally produced by loading the greater trochanter it was frequently found that the trochanter itself was undamaged.

Anatomically the greater trochanter consists of a thin shell of compact bone filled with spongy bone and the fact that it can support a load, sometimes more than 1000 lbs. sufficiently great to fracture the bone without the trochanter itself being damaged is of considerable interest. In x-rays of clinical fractures of the femoral neck and trochanteric region often arising from falls on the trochanter the trochanter frequently appears to be undamaged. In the living body the soft tissue overlying the greater trochanter would absorb much of the energy of a fall but such was not the case in the femoral fractures produced in dry bones by static or dynamic loading of the greater trochanter. In these cases the energy was very likely absorbed by the spongy bone itself which is a good energy absorbing material.

Shearing Strength of Bone

The direct shearing strength of bone was first tested by Rauber (*loc. cit.*) who designed a special tool which subjected standardized specimens of compact and spongy bone to a double shearing action. Rauber emphasized that in shearing tests bending can only occur in two directions but if tested in the third

direction, parallel with the fibers, great differences are found Consequently, some of his samples were cut parallel with the long axis of the bone and loaded perpendicular to the direction of the fibers while others were cut perpendicular to the long axis of the bone and tested parallel with the direction of the fibers

Rauber found that the average shearing strength of five fresh rods of compact bone, cut parallel to the long axis of the femur of a man 30 years of age and loaded perpendicular to the direction of the fibers was 16850 lbs/in^2 (14931-18486) The average shearing strength of five other rods of compact bone cut from the same femur perpendicular to its long axis and loaded parallel with the fibers was only 7152 lbs/in (6043 9243)

Six fresh rods of spongy bone, cut parallel with the long axis of a tibia but loaded perpendicular to the long axis, had an average shearing strength of 796 lbs/in (644 888) Five other fresh samples of spongy bone from the same tibia cut perpendicular to its long axis and loaded parallel with it had an average shearing strength of 270 lbs/in^2 (199 419)

Rauber s tests clearly show that the shearing strength of compact and spongy bone is considerably greater when loaded perpendicular to the direction of the fibers or the long axis of the bone than when loaded parallel with the fibers or the long axis of the intact bone Therefore when figures for the shearing strength of bone are given the direction of the shearing force with respect to the fibers or long axis of the bone should also be included

Differences in the shearing strength of specimens from various parts of the same bone as well as the influence of moisture on the shearing strength of bone have been studied by Evans and Lebow (1951) in 242 samples of compact bone from the anterior posterior lateral and medial quadrants of the proximal middle and distal thirds of the femoral shaft The samples were cut parallel with the long axis of the bone and machined to a standardized size Of the two samples obtained from each region one was air dried and tested dry while the other was kept in water and tested wet A special shearing tool was designed to hold the samples

whose ultimate shearing strength was tested in a 5000 lb capacity testing machine calibrated to an accuracy of ± 1%. The shearing load was applied perpendicular to the long axis of the bone and the direction of the fibers.

The greatest average shearing strength, with respect to the third of the bone from which the samples were obtained, was 8320 lbs/in^2 for the dry samples from the proximal third of the bone and 10840 lbs/in^2 for the wet samples from the middle third of the bone. The samples from the distal third of the shaft were the weakest, the average shearing strength being 7770 lbs/in^2 and 9350 lbs/in^2 for the dry and wet samples, respectively.

With respect to the quadrant of the bone from which the samples were obtained the lateral was the strongest (8880 lbs/in^2) for the dry samples and the medial (10190 lbs/in^2) for the wet samples. The weakest samples, both dry (7230 lbs/in^2) and wet (9410 lbs/in^2), were from the anterior quadrant.

The average shearing strength of all the dry samples was 8000 lbs/in^2 and of all the wet samples 9800 lbs/in^2. The average ultimate tensile strength for similar samples from the femoral shaft was 15330 lbs/in^2 and 11840 lbs/in^2 for dry and wet specimens, respectively. Thus, air drying decreased the shearing strength but increased the tensile strength of the samples. However, the tensile samples were tested parallel with the long axis of the bone and the fibers instead of perpendicular to them as were the shearing specimens.

Summary

Studies of the strength characteristics of compact bone give the following results. Compact bone, when loaded parallel with the long axis of the bone or the collagen fibers, is strongest in compression, intermediate in tension, and weakest in shear. Drying increases its compressive and tensile strength but reduces its shearing strength. Its tensile and compressive strength are decreased by heating. The compressive strength is greater when loaded parallel with the long axis of the bone or the direction of its fibers than when loaded perpendicular to them. Embalming

tends to reduce the tensile and compressive strength of compact bone The average ultimate tensile and compressive strength of human compacta is similar to that of other mammals

The compressive strength of spongy bone is considerably less than that of compact bone and is approximately similar in man and other mammals The tensile and shearing strength of spongy bone have not been determined

The stress strain curve for dry compact bone under tension is approximately a straight line to failure but in wet bone the curve deviates from a straight line as the failure point is approached The latter is also true for the stress-strain curve of wet compact bone under compression The per cent elongation or strain in wet and dry compact bone under tension is directly proportional to its energy absorbing capacity However wet bone has a greater energy absorbing capacity than has dry bone

15

Torsion, Bending and Fatigue Strength of Bone

In bone as in other materials, torsion produces tensile and shearing stresses while bending gives rise to tensile and compressive stresses on the convex and concave sides respectively, of the test specimen However the results of determinations of the bending and torsion strength of bone are not directly comparable with those for tension compression and direct shear because the former are computed on the assumption that the modulus of elasticity is a constant to rupture This as indicated by the stress strain curve is not true of bone especially when it is wet or has a relatively large amount of water as is the case in living bone Torsion is measured in inch pounds or kilogram meters

The severity of bending is measured by the bending moment applied to the specimen If the specimens have constant dimensions the bending moment is directly proportional to the load applied to the specimens and their strength may be compared on the basis of the maximum load they supported Sometimes the extreme fiber stress (lbs/in² or kg/mm²) is computed but this is not comparable with stress values for direct tension or compression because the modulus of elasticity is not constant

Torsion Strength of Bone

The torsion strength of little rods of fresh compact bone from the femur of a man 30 years of age was determined by Rauber (*loc cit*) The samples had a reduced circular area 80 mm long 1 93 mm in diameter and expanded rectangular ends for gripping purposes The load was applied via a wheel with a 10 cm radius The breaking torque for the four samples tested was 1000 1250 1000 and 1100 gm cm The average torsional

shearing strength was 11233 lbs/in² which is quite similar to that 10633 ± 1140 lbs/in², found by Dempster and Liddicoat (*op. cit.*) for dry spindles of compact bone from six human tibias.

Rauber found that the torsion strength of compact bone is greater than the direct shearing strength of samples cut perpendicular to the long axis of the bone and loaded parallel with the direction of the fibers. However, the direct shearing strength of samples cut parallel with the long axis of the bone but loaded perpendicular to the fiber direction is greater than the torsional shearing strength. Dempster and Liddicoat reported that the average torsional shearing strength of compact bone is considerably less than its compressive strength both for wet and dry bone. The torsional shearing strength is also less than the average tensile strength for dry bone but there is some overlap in the range of variation of the tensile strength of wet bone and the torsional shearing strength of dry spindles.

The torque necessary to produce torsion fracture of intact bones was first determined by Messerer (1880), who recorded the following average loads applied at the end of a lever arm 16 cm. long for fresh human bones: clavicle—8 kg.; humerus—40 kg.; radius—12 kg.; ulna—8 kg.; femur—89 kg.; tibia—48 kg. and fibula—6 kg. The unit ultimate shearing strength in torsion for the femurs of a man 29 years of age was 810 and 824 lbs/in². The average load required to fracture 12 femurs was 1350 inch lbs. of torque (694–1971 inch lbs.). Pedersen, Evans and Lissner (1949) produced fracture of four dry embalmed femurs with an average of 348 [illegible] inch lbs. of torque (166–498), while Carothers, Smith and Calabrisi (1949) reported an average of 449 inch lbs. (142–827) for three similar femurs. Comparison of the figures obtained for fresh and embalmed femurs seems to indicate that embalming reduces the strength of bone in torsion.

In contrast to the other investigators of torsion in intact bones, Carothers, Smith and Calabrisi (1949) plotted torque-angle of twist curves for one bone where a slight deviation from a straight line occurred at a high torque. This bone failed at an ultimate torque of 827 inch lbs. However, only three femurs were used so no generalized conclusions can be drawn.

Bending Strength

The first experiments on the bending strength of bone were made by Rauber (*loc. cit.*) who used little rods of compact bone with a cross section area of 4 mm. The specimens were obtained from fresh and dry unembalmed human bones and were loaded to failure in an apparatus designed for bending tests. The results he obtained are presented in Table X.

TABLE X

BREAKING LOAD OF INTACT HUMAN BONES

(Data from Rauber, 1876)

Source	*Test Conditions*	*Ultimate Bending Load (gms.)*	*No. of Samples*
Humerus 30 yr. male	Dry in 15–25° C	1265 (1200–1330)	2
Femur 30 yr. male	Dry in 15–25° C	2375 (2100–2650)	2
Femur 70 yr. male	Dry in 15–25° C	1166 (1350–1550)	3
Femur 46 yr. male	Dry in 15–25° C	2537 (1000–4000)	4
Femur 46 yr. male	Fresh in 15–25° C	1800	1
Tibia 46 yr. male	Fresh in 15–25° C	2150 (2100–2200)	2
Tibia 46 yr. male	Fresh in 38° C	1350 (900–1800)	2
Femur 46 yr. male	Fresh in 38° C	1866 (1600–2200)	[illegible]
Femur 46 yr. male	Dry, continuous load	1200 (1100–1300)	2
Femur, ox	Dry in 15–25° C	1966 (1600–2600)	4
Femur, ox	Fresh in 38° C	2507 (2000–3250)	7

Rauber's results reveal that the average ultimate bending load and its range of variation are similar for human and ox bones. The bending load a bone can support without breaking is reduced by drying the bone or heating it. The few tests made indicate that fresh ox bones can support a greater bending load before failure than can dry ones. However, more data are needed before definite conclusions can be drawn.

The load required to fracture small rectangular specimens of compact bone from man and other mammals has been the subject

of an extensive series of studies by the Italian investigators Maj Olivo and Toajari. Specimens of a standardized size were tested under bending in an apparatus called a clasimeter which consists of two knife blades 7 mm apart, for supporting the specimen while the load is applied to it midway between the supports. The fracturing load was recorded in kilograms and the deflection in millimeters. The loading was in the direction of the thickness or smallest dimension of the specimen. Variations in the load required for fracturing specimens from the same section of a single bone, as well as from one bone to another, were studied with respect to the histological structure of the specimens.

Maj and Toajari (1937) tested the breaking load of 12 little parallelepipeds from ox tibias cut parallel, radial and tangential to the long axis of the bone. The stereometric position of the

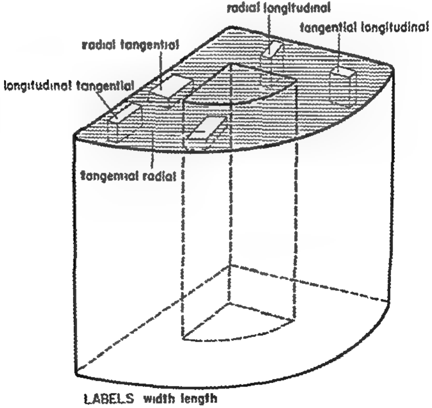

Figure 41. Stereometric position of specimens tested by Maj & Toajari (1937).

specimens in the intact bone (Figure 41) was described by two terms, the first indicating the orientation of the width of the specimen and the second the orientation of the length of the specimen with respect to the long axis of the bone.

TABLE XI

Breaking Load Under Bending of Specimens of Compact Bone from Ox Limbs

(Data from Maj and Toajari 1937)

Orientation of Specimen	Specimen Dimensions in mm: Width	Thickness	Fracture Load (kg) for a Span of [illegible] mm			
			Exp I	*Exp II*		
Radial longitudinal	2.9	1.6	18.75	19.00		
Radial tangential	2.9	1.6	8.50	8.50		
Tangential radial	2.9	1.6	4.25	4.26		
			Exp III	*Exp IV*		
Radial longitudinal	3.55	1.9	13.50	12.00		
Radial tangential	3.55	1.9	19.00	19.00		
Tangential radial	3.55	1.9	9.00	9.00		
			Exp V	*Exp VI*		
Radial longitudinal	2.55	2.55	Not broken by 45 kg			
Radial tangential	2.55	2.55	23.50	23.50		
Tangential radial	2.55	2.55	11.25	12.00		
			Exp VII	*Exp VIII*		
Radial longitudinal	4.1	2.05	Not broken by 45 kg			
Radial tangential	4.1	2.05	22.00	21.75		
Tangential radial	4.1	2.05	11.25	11.25		
			Exp IX	*Exp X*	*Exp XI*	*Avg*
Radial longitudinal	2.65	1.2	14.25	15.50	14.50	15.00
Tangential longitudinal	2.65	1.2	16.00	15.00	15.25	
Longitudinal tangential	2.65	1.2	4.50	5.25	4.50	5.00
Radial tangential	2.65	1.2	5.25	4.50	5.25	
Longitudinal radial	2.65	1.2	2.50	3.00	2.75	2.50
Tangential radial	2.65	1.2	2.50	2.00	2.50	

The results of the tests (Table XI) revealed that the load required to fracture the specimens cut longitudinally to the long axis of the bone was three times greater than that for specimens cut tangentially to the axis and six times greater than that of specimens cut radially to the axis. Furthermore, the tangentially cut specimens were about twice as strong as the radially cut specimens. The longitudinally cut specimens were thus seen to be the strongest and the radially cut ones the weakest as far as

of an extensive series of studies by the Italian investigators Maj, Olivo and Toajari. Specimens of a standardized size were tested under bending in an apparatus called a clasimeter which consists of two knife blades, 7 mm. apart, for supporting the specimen while the load is applied to it midway between the supports. The fracturing load was recorded in kilograms and the deflection in millimeters. The loading was in the direction of the thickness or smallest dimension of the specimen. Variations in the load required for fracturing specimens from the same section of a single bone, as well as from one bone to another, were studied with respect to the histological structure of the specimens.

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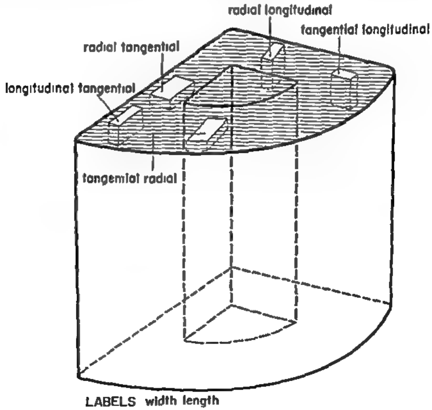

Figure 41. Stereometric position of specimens tested by Maj & Toajari (1937).

which were the weakest few of the collagen fibers were parallel to the major axis of the specimens. Thus the greater resistance to bending in a specimen of compact bone appears to be correlated with the greater number of collagen fibers parallel to the major axis of the specimen. On the basis of the results of their tests Maj and Toajari made the following conclusions: (1) the resistance of compact bone to bending failure is directly proportional to the number of collagen fibers present in the plane of the section of the bone, (2) the cohesiveness of the interfibrillar calcified substance is at least six times inferior to that of the collagen fibers, (3) the characteristic mechanical antitropism of bone is the result of the distribution and direction of the collagen fibers, and (4) the interfibrillar substance very probably confers homogeneity and isotropic properties to compact bone.

Olivo, Maj and Toajari (1937) using the clasimeter determined the breaking load (kg.) and reported the deflection in hundredths of a millimeter per 5 kg. load of several specimens from the same ring of bone taken from the metacarpals and metatarsals of an ox six years of age, a mule, and the femurs of two women 42 and 90 years of age. The number and orientation within the bony rings of the various specimens tested are illustrated in Figure 42.

The results of the tests of the metatarsal specimens showed a definite tendency toward an increase in the anteroposterior direction, in the breaking load and a decrease in the magnitude of the deflection. Such orientation in the lateromedial direction was not observed.

The fracturing load and deflection in a series of specimens taken from the anterior and posterior part of the femur of a man 79 years of age are summarized in Table XII. No consistent positive relation was evident between the breaking load and deflection, the same load producing varying amounts of deflection in different specimens. The average breaking load was 8.0 kg. for the specimens from the anterior part of the femur and 8.4 kg. for those from the posterior part. The difference between the average breaking load of specimens from the anterior and posterior part of the femur was less than 1% and probably is not statistically significant.

the load necessary to fracture them by bending was concerned

From histological examination of the specimens under normal and polarized light the differences in the magnitude of the fracturing load was explained on the basis of the predominate direction of the collagen fibers in the individual test specimens In the longitudinal specimens, which were the strongest the predominate direction of the collagen fibers was parallel with the major axis of the specimens However, in the radial specimens

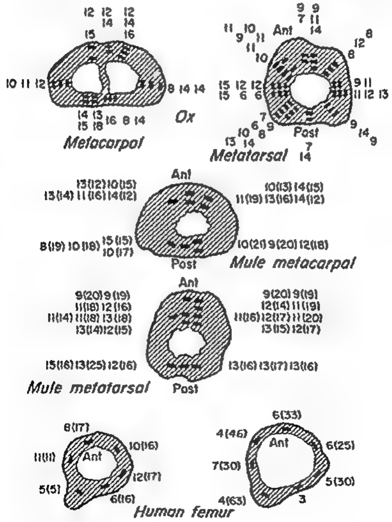

Figure 42 Breaking load (kgs) and deflection 0 01 mm/5 kg load (in parentheses) (Redrawn from Olivo Maj and Toajari Bull sc med Bologna 109 1937)

ing by use of the clinometer. Considerable regional variation (Table XIII) was found in the breaking load of specimens from different rings of a single bone. The specimens from the ring in the middle of the diaphysis were the strongest, while those from rings on either side of it showed a progressive decrease in the breaking load. The decrease was rather gradual from the center of the diaphysis toward the proximal end, whereas a sharp drop occurred from the middle toward the distal end of the bone.

TABLE XIII

REGIONAL DIFFERENCES IN AVERAGE BREAKING LOAD (KG) UNDER BENDING IN OX COMPACT BONE

(Data from May 1938)

Bone	*Distance (cm) from Proximal End of Intact Bone*	*Medial Quadrant*	*Lateral Quadrant*	*Anterior Quadrant*	*Posterior Quadrant*
Metatarsal	7	7 50	9 50	7 25	5 00
	12	11 00	10 00	10 00	7 50
	17	11 20	11 00	10 30	10 00
	22	1 50	1 50	9 50	8 75
	27	6 00	4 00	6 50	1 50
Metacarpal	5	10 [illegible]	9 50	9 00	6 00
	8	10 50	9 50	9 00	[illegible] 00
	10	10 50	11 00	10 00	9 00
	15	11 00	12 00	13 50	14 00
	20	4 00	3 00	4 75	3 50

Differences were also found (Table XIV) within various quadrants of a single bony ring. The quadrants in order of decreasing breaking load of the specimens taken from them were grouped as medial, lateral, anterior and posterior. One exception was found in the metacarpal ring 15 cm from the proximal end of the bone in which the breaking load was least in the specimens from the medial quadrant and greatest in those from the posterior quadrant. The rings nearest the distal end of the bone were also exceptions because the breaking load of the specimens from the anterior and medial quadrants was slightly greater than that of the specimens from the posterior and lateral quadrants.

Even within a single quadrant (Table XIV) differences were found in the breaking strength of the specimens. Thus the breaking strength of the specimens decreased from the external surface toward the medullary cavity in the posterior, medial and lateral quadrants of the metatarsal and the same tendency was

TABLE XII

BREAKING LOAD AND DEFLECTIONS IN SPECIMENS OF COMPACT BONE FROM THE FEMUR OF A MAN 79 YEARS OF AGE

(From Olivo, May and Tonjari, 1937)

	Anterior Part													
Distance (cm) from superior end of bone	8	10	12	14	16	18	20	22	24	26	28	30	32	34
Breaking load (kg.)	11	7	7	9	[illegible]	7	8	6	11	9	7	8	7	6
Deflection (mm.) per 5 kg. load	20	20	19	17	19	20	20	22	16	22	22	21	22	30
	Posterior Part													
Breaking load (kg.)	4	7	7	7	7	7	8	8	20	9	[illegible]	9	9	7
Deflection (mm.) per 5 kg. load	32	22	20	23	16	21	21	22	20	17	17	18	18	20

Olivo (1937) reported the existence of a constant parallelism between the mechanical behavior of bone specimens and varying microscopic structures. His work was based on studying various areas of metacarpals and metatarsals of the ox, the horse, the dog and especially the chamois. In areas of the bone showing marked fracture resistance and a high modulus of elasticity there was a predominance of osteones composed of collagen fibers having a vertical or steeply spiralling course. In the areas with a low resistance to fracture the fibers had a more circular or oblique course. Olivo did not, however, report the actual values for the breaking load of the specimens.

May (1938) studied topographic differences in the mechanical strength of specimens of compact bone from the same skeletal segment. Complete diaphyses of oxen metatarsals and metacarpals and of human femurs were used. From the metatarsal of a six year old ox five bony rings 2 cm. thick were sawed at distances of 7, 12, 17, 22, and 27 cm., respectively, from the proximal end of the bone. From the metacarpal of the same animal five rings of similar thickness were cut at distances of 5, 8, 10, 15, and 20 cm. from the proximal end of the bone.

From the four quadrants of each ring test specimens 1.2 mm. thick, 3 mm. wide and 10–20 mm. long, each oriented with respect to the anatomical axis of the diaphysis, were cut. The dimensions of the specimens were calibrated to hundredths of a millimeter. The specimens were then tested to failure under bend

TABLE XX

COMPARISON OF THE BREAKING LOAD (BENDING) AND POROSITY IN THE FEMUR OF A MAN 70 YEARS OF AGE

(From May 1938)

Distance from the Proximal End (cm.) of the Bone	8	10	12	14	16	18	20	22	24	26	28	30	32	34
Anterior Aspect of the Shaft. Breaking Load in kg	11	7	7	9	9	7	8	6	11	9	[illegible]	8	[illegible]	[illegible]
Per cent Porosity	14.6	12.2	10.0	10.1	9.25	8.02	8.25	12.2	10.4	9.60	14.1	23.0	17.1	23.3
Posterior Aspect of Shaft Breaking Load in kg	4	7	7	7	7	7	8	8	10	[illegible]	9	9	9	7
Per cent Porosity	16.7	10.2	14.1	10.4	9.70	10.8	8.34	10.2	6.86	6.10	9.10	14.1	9.25	12.[illegible]

TABLE XIV

REGIONAL DIFFERENCES IN THE AVERAGE BREAKING LOAD (KG) UNDER BENDING OF OX COMPACT BONE

(Data from May 1938)

Bone	Distance (cm) of Specimen From Proximal End of Bone	Anterior Quadrant			Posterior Quadrant			Lateral Quadrant			Medial Quadrant		
		Ext	Mid	Int	Ext	Mid	Int	Ext	Mid	Int	Ext	Mid	Int
Metatarsal	7	6	—	8 5	5	—	—	9	—	10	9	—	6
	12	10 5	11 5	8	8	—	7	13	—	7	13	—	9
	17	10 5	10 5	10 5	11 5	7	8	15	9	9	13	11 5	10
	22	10	—	9	10	—	7 5	9	—	10	8	—	11
	27	6 5	—	—	4 5	—	—	4	—	—	6	—	—
Metacarpal	5	7	—	9 5	6	—	—	11	—	8	12	—	9
	8	6	—	10 5	7	—	9	12	—	7	12	—	9
	10	8 5	—	11 5	11	—	7	11	13	9	11	7	10
	15	12	14	15 5	16 5	12 6	13 5	14	14	8	10	11	12
	20	4 75	—	—	3 5	—	—	3	—	—	4	—	—

in Table XVI. The results are presented as a fraction the numerator of which indicates the breaking load in kilograms while the denominator indicates the deformation for a length of 7 mm in hundredths of a millimeter per 5 kg. of load. These specimens

TABLE XVI

Average Breaking Load in Kg. (Numerator) and Deformation in Hundredths of a Mm. [illegible] Kg. Load [illegible] a 7 Mm. Span (Denominator) in Compact Bone from [illegible] Mule & Ox

Data from [illegible]

Metacarpal

Animal	Site of Sample	Zone	Anterior Section	Posterior Section
Mule	12-14 cm from proximal end of bone	External	11 [illegible] kg. (10-14) / 14 [illegible] mm (12-18)	11 (10-12) / 17 (17-18)
		Middle		[illegible] 2[illegible] (8-10) / 19 [illegible] (18-21)
		Internal	13 (11-14) / 14 (12-16)	11 [illegible] (8-1[illegible]) / 1[illegible] [illegible] (14-17)
	18-20 cm from proximal end of bone		10 8 ([illegible]-13) / 1[illegible] 7 (13-17)	8 [illegible] (6-10) / 17 12[illegible] (13-19)
Ox	Middle of diaphysis	External	8 12[illegible] ([illegible]-10) / 17 [illegible] (14-20)	8 88 (8-13) / 18 63 (17-20)
		Internal	8 47 (6-12) / 18 48 (16-22)	10 2[illegible] ([illegible]-13) / 17 63 (16-19)

Metatarsal

Animal	Site of Sample	Zone	Anterior Section	Posterior Section
Mule	8-10 cm from proximal end of bone	External	9 (9) / 20 (19-22)	
		Middle	11 [illegible] (11-13) / 16 [illegible] (14-20)	13 (12-1[illegible]) / 17 66 (16-2[illegible])
		Internal	12 [illegible] (12-13) / 14 [illegible] (13-17)	
	15-17 cm from proximal end of bone	External	10 [illegible] (9-11) / 18 16 (12-18)	11 2[illegible] (13-1[illegible]) / 16 [illegible] (14-19)
		Middle	12 7[illegible] (12-14) / 17 2[illegible] (1[illegible]-19)	
		Internal	11 [illegible] (13-1[illegible]) / 1[illegible] 6[illegible] (14-16)	14 3 (14-15) / 14 3 (14-15)
Ox	Middle of diaphysis	External	8 0 (7-10) / 18 4 (17-21)	10 7[illegible] (8-13) / 22 0 (18-27)
		Middle	[illegible] 0 (7-12) / 18 0 (1[illegible]-20)	11 5 (10-13) / 17 7[illegible] (13-21)
		Internal	[illegible] 2[illegible] (6-10) / 18 [illegible] (17-21)	10 8 (10-12) / 16 0 (1[illegible]-19)

generally seen in the metacarpal specimens In the anterior quadrant of both the metatarsal and the metacarpal the opposite tendency was noted

The relation between fracture load and porosity of specimens from the anterior and posterior aspects of the femur of a man 79 years of age was also studied The specimens were taken at 2 cm intervals from strips of bone cut the length of the femur The porosity was determined by computing the volume occupied by the cavities of the specimens The metatarsal of an ox was similarly analyzed

The results of the study (Table XV) showed that the breaking load decreased and the porosity increased from the approximate middle of the femur toward its extremities This was particularly true of the breaking load for specimens from the posterior aspect of the shaft However, the increase in the porosity of the specimens was not proportional to the decrease in the breaking load

From his studies Maj concluded that the degree of porosity was not responsible for the variations in breaking load except in the distal metacarpal rings in which the porosity was approximately 50% Consequently according to him the differences in the strength of compact bone of different areas of a single skeletal segment are probably a function of intrinsic properties of the bone and variations in the density and orientation of the collagen fibers

The relation of breaking strength and elasticity of compact bone to its histological structure was also studied by Toajari (1938) For this purpose he selected bones which because of their peculiar position and function he considered to be continuously subjected to a particular mechanical stress Test specimens 1 2 mm thick and 3 mm wide were cut parallel with the long axis of the bone from rings of bone taken from the radius the ulna and the olecranon processes of oxen and the metacarpals and metatarsals of oxen horses and mules The specimens were progressively loaded to failure in the clasimeter

The data Toajari obtained on the breaking load and deflection produced over a span of 7 mm in specimens from the metatarsal and metacarpal of adult mules and oxen are summarized

In order to verify some of his conclusions Toyama (1935) examined splinters from fracture specimens under polarized light. He found that in longitudinal sections the luminosity was significantly greater than in transverse sections because of the fact that the collagen fibers usually run longitudinal with respect to the major axis of the bone. In comparing transverse sections of the strongest and the weakest specimens he found more pronounced luminosity in the latter. However, in comparing longitudinal sections the luminosity in the specimens showing the greatest fracture strength (15 kg.) was greater than those showing the least fracture strength (9 kg.).

TABLE XVII

BREAKING LOAD AND CAVITY VOLUME IN METACARPALS OF TWO BREEDS OF CATTLE

(Data from Toyama 1935)

	Samples from Different Parts of Bone				Average Breaking Load (kg.)	Samples from Different Parts of Bone				Average Breaking Load (kg.)
	1	2	3	4		1	2	3	4	
Dutch Cows										
I	3	9	3	—	5.0	7	7	11	—	8.3
II	9	7	8	—	8.0	7	6	7	—	6.7
III	4	3	8	—	5.0	4	8	4	—	5.3
IV	4	5	7	6	5.5	10	6	11	—	9.0
					5.9					7.3
Jersey Cows										
I	10	9	9	4	8.0	9	7	5	8	7.2
II	7	7	6	4	6.0	11	7	7	7	8.0
III	6	8	8	7	7.2	9	10	7	9	8.7
IV	7	6	6	7	6.7	10	9	8	7	8.5
					7.0					8.1
					Average Cavity Volume (%)					*Average Cavity Volume (%)*
Dutch Cows										
I	4.1	3.4	3.8		3.8	7.4	4.3	4.3	—	5.3
III	4.3	3.3	2.8		3.5	4.4	5.1	7.3	—	5.6
					3.6					5.5
Jersey Cows										
I	3.1	2.4	2.6	—	2.6	3.7	2.6	3.1	—	3.1
III	1.6	2.2	1.9	—	1.9	4.2	3.7	4.3	—	4.1
					2.3					3.6

were obtained from different depths of the diaphysis of the bone, i e external, middle and internal zones the external being nearest the periphery of the bone and the internal nearest the medullary cavity

From the 271 observations Tojan stated that in general the modulus of elasticity increases with the increase in the breaking strength of the bone He felt that the parallel behavior of these two physical properties was justified on the basis that the mechanical properties are dependent upon the quality and orientation of the collagen fibers present in the bony tissue In a detailed analysis many differences were found in the histological structure of the same skeletal segment as far as its breaking strength was concerned This was particularly true in the metatarsals and the olecranon process in which the areas or zones of bone usually subjected to tension had a more marked resistance to fracture and a relatively higher modulus of elasticity than the areas subjected to compression, in which the breaking strength and modulus of elasticity were greatly reduced This was true in quite a large percentage of cases

Toajan s finding regarding the correlation between resistance to fracture modulus of elasticity and areas of a bone subjected to tensile or compressive stress was partially confirmed by Evans and Lebow (1951) in their studies on the human femur In the erect posture the lateral aspect of the femur is primarily subjected to tensile stress while the medial aspect is under compression Because of the forward bowing of the femoral shaft the anterior and posterior aspect of the diaphysis are partly subjected to tensile and compressive stress respectively Evans and Lebow found that the average tensile strength of specimens from the lateral quadrant of the shaft was the greatest for the entire diaphysis but the same specimens did not have the highest modulus of elasticity The specimens from the anterior quadrant of the shaft had the highest modulus of elasticity and the lowest tensile strength However the modulus of elasticity of the specimens from the lateral quadrant the region of greatest tensile stress and strain in the erect posture was slightly greater than that of specimens from the medial quadrant the area of greatest compressive stress

fibers per square millimeter of bone and the average area of the collagen fibers in square micra. Unfortunately, however, he did not record from which half, anterior or posterior, of the metacarpal his material was taken. Consequently, it is impossible to determine from his data what relationship, if any, exists between the breaking load and the number and size of the collagen fibers.

From Table XVII it is seen that the average breaking strength of both halves of the metacarpal bones of the Polish cows was greater than that for the Dutch cows. However, the percentage volume of the cavities, that is, the porosity, of both halves of the metacarpals of the first and third Dutch cows was greater than corresponding values for the metacarpals from the first and third Polish cows. Thus the degree of porosity and the breaking strength seem to be parallel, just the opposite to what was found in the anterior and posterior halves of the individual metacarpals in the two races of cattle.

The greater bending strength of the metacarpals of the Polish cattle (Table XVIII) may be correlated with the fact that the percentage area occupied by collagen fibers, as well as the average area of the fibers themselves (in terms of square micra), were greater in the metacarpals of the Polish cattle than in those of the Dutch cattle. However, the number of collagen fibers per square millimeter was greater in each of the metacarpals of the Dutch cattle than in those of the Polish cattle.

The influence of prolonged x-ray therapy on the breaking strength of dog bone was also investigated by May (1940). A large adult dog weighing 15 kg. was subjected to two separate cycles of x-ray irradiation, one on the external portion of the left thigh corresponding to the site of the diaphysis of the femur and the other on the left forearm. The left thigh was irradiated with a dosage of 600r three times at intervals of one month, giving a total dosage of 1800r. The left forearm was irradiated alternately on the medial and the lateral aspects with doses of 200r repeated 40 times with an interval of one day between every two treatments, resulting in a total dosage of 8000r. The animal remained in optimum organic condition during the course of the experiment and radiographs taken 45, 60, and 150 days after the start

Differences in the histological structure and breaking strength of compact bone from two races of domestic cows, one from Holland and the other from Poland were studied by Toajari (1939) Test pieces 3 mm by 1 2 mm were taken from the anterior and the posterior regions of the metacarpals of the cattle The pieces were tested to failure under bending in the clasimeter the load being recorded in kilograms Whenever possible four samples from each half of each metacarpal were tested

The average breaking load (Table XVII) was greater in the posterior half of all metacarpals in both races of cows However, some overlapping was noted in the breaking load of individual specimens

The porosity of the bone in terms of the percentage area of the cavities i e Haversian canals and Volkmann s canals was also greater in the posterior half of the metacarpals than in the anterior half Thus, (Table XVII) the stronger half of the bones as far as the breaking load is concerned also had the highest percentage volume of space or were the more porous This relationship between the average breaking load of the region and its porosity suggests that the quality of the bone in the posterior half of the metacarpals is better that is stronger per unit area than that in the anterior half of the metacarpals The unit strength of the specimens was not determined

Toajari (Table XVIII) computed the per cent of the area of the bone occupied by the collagen fibers the numbers of collagen

TABLE XVIII

NUMBER AND SIZE OF COLLAGEN FIBERS IN METACARPAL BONES OF TWO BREEDS OF CATTLE

(*Data from Toajari 1939*)

Dutch Cows	*Per Cent of the Area of Bony Tissue Occupied by Collagen Fibers*	*Number of Collagen Fibers per mm² of Bone*	*Average Area of Section of Collagen Fiber in sq micra*
I	6 7	180 411	0 374
II	9 0	247 021	0 360
III	14 1	272 004	0 511
IV	11 2	266 902	0 410
Polish Cows			
I	30 6	147 104	2 046
II	24 6	102 695	2 338
III	15 7	152 695	1 517
IV	20 1	99 920	1 988

fibers per square millimeter of bone and the average area of the collagen fibers in square micra. Unfortunately, however, he did not record from which half, anterior or posterior, of the metacarpal his material was taken. Consequently, it is impossible to determine from his data what relationship, if any, exists between the breaking load and the number and size of the collagen fibers.

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The influence of prolonged x-ray therapy on the breaking strength of dog bone was also investigated by Maj (1940). A large adult dog weighing 15 kg was subjected to two separate cycles of x-ray irradiation, one on the external portion of the left thigh corresponding to the site of the diaphysis of the femur and the other on the left forearm. The left thigh was irradiated with a dosage of 600r three times at intervals of one month giving a total dosage of 1800r. The left forearm was irradiated alternately on the medial and the lateral aspects with doses of 200r repeated 40 times with an interval of one day between every two treatments, resulting in a total dosage of 8000r. The animal remained in optimum organic condition during the course of the experiment and radiographs taken 45, 60, and 150 days after the start

of the irradiation showed no difference between the irradiated and the nonirradiated bones

Three months after the end of the roentgen treatments the animal was sacrificed and autopsied No alteration in the macroscopic aspect of the irradiated muscles or bones was found and both the irradiated and the nonirradiated bones had the central cavities filled with yellow marrow From the femoral and radial diaphysis of the irradiated and nonirradiated limbs seven six and nine rings each 2 cm thick, were cut The rings from the center of the diaphysis were submitted to controlled histological examination to see whether irradiation had modified the architecture of the bone with respect to the Haversian systems lamellae and interstitial substance Such an examination revealed no difference between the irradiated and normal bone

From the anterior, posterior and lateral aspects of the femoral rings specimens of bone 1 2 mm thick 3 mm wide and 10 to 20 mm long were tested to failure in the clasimeter The breaking load of the specimens from the irradiated and normal femur was practically identical being 10 3 kg and 10 4 kg respectively However the specimens from the radius showed a decrease in breaking load of the irradiated bones which had an average of 6 5 kg compared with 9 6 kg for the nonirradiated bones The same was true of the ulna the irradiated specimens having an average breaking load of 10 6 kg compared with 12 3 kg for the nonirradiated specimens From his studies Maj concluded that irradiation reduced the bending strength of the radius more than the ulna both of which were subjected to more intense radiation than the femur

The calcium and phosphorus content of fractures of irradiated and nonirradiated bones were then studied in an attempt to learn if the reduced fracture strength of the irradiated bones resulted from resorption of calcium salts or a chemical change The chemical analysis showed that irradiation had no effect whatever on the quantity of the inorganic salts present nor on the chemical composition of bone Therefore according to Maj the causes of minor variations in the strength of irradiated bone must be sought in (1) the modification and dispersion of particles constituting the mineral substance (2) difference in the intrinsic prop

erties of osteocytes and the fundamental substance (osteomucoid and collagen fibers), or (3) in a combination of these factors. The theory that irradiation causes variations in dispersion patterns and in crystalline states in aggregations was considered. However no definite conclusions were drawn and Maj believes that modifications in the strength of bone induced by x-ray irradiation requires further study.

In a later paper Maj (1942) studied regional variation in the bending strength of human bones with respect to the age of the individual from whom the bones had been obtained. Rings about 2 cm thick were cut from the middle of the diaphysis of the humerus, ulna, femur, and tibia of 10 individuals varying from five to 90 years of age. From five of them the intermediate segments of the humerus, femur and tibia were removed for ease of sectioning. From the anterior, posterior, medial and lateral quadrants of each ring test specimens of bone 1.2 mm thick, 3 mm wide and 10 - 15 mm long, all oriented in the direction of the major axis of the diaphysis, were prepared. Six trial specimens from the humerus, three from the ulna, nine from the femur and six from the tibia were prepared. The breaking load by bending was tested with the clasimeter. The flexion of the specimens was measured by a beam of reflected light which magnified it 100 times.

In the first orienting observations variations of 5, 6, and 7 kg and in a few cases as high as 8 or 9 kg were noted between one reading and another. This heterogeneity was more evident in human bones than those of lower animals and was greater and more marked in the femur and humerus than in the tibia and ulna. The average breaking loads for the various specimens were the ulna—8.36 kg, the tibia—7.7 kg, the humerus—7.17 kg, and the femur—6.86 kgs.

In rings from specific zones of the diaphysis the maximum average values for bending strength were distributed with certain regularity and had a somewhat constant relationship, but the minimal values were very inconsistently distributed, and for the femur, the humerus, and to a lesser extent the tibia, it was sometimes difficult to determine which sections of the bone were the stronger.

TABLE XIX

REGIONAL DIFFERENCE IN BREAKING LOAD (KG) UNDER BENDING OF HUMAN COMPACT BONE

(*From May 1942*)

Bone	*Anterior Quadrant*	*Medial Quadrant*	*Posterior Quadrant*	*Lateral Quadrant*
Humerus	7 51	6 60	7 37	7 36
Ulna	9 38	8 87	7 17	—
Femur	6 92	6 91	6 02	7 10
Tibia	7 30	7 91	7 26	7 46

However on the basis of average values for breaking loads obtained from all the tests, May deduced that the physical heterogeneity of the bone was not a chance matter In some cases certain areas of the bone were consistently and logically more resistant than others (Table XIX) He, therefore affirmed that

(1) if the anterior posterior and lateral quadrants of the humerus show minimal differences in the average breaking strength of the bone, the medial quadrant is logically weaker than the others (2) in the femur the bending strength diminishes markedly in the posterior quadrant in the region of the linea aspera and (3) in the tibia the medial quadrant has the highest bending strength In the ulna the anterior quadrant had the greatest bending strength and the posterior the least bending strength

With respect to the age of the individual from whom the specimens were obtained the following conclusions were drawn (1) the highest fracture loads were found in the greatest statistical percentage in two periods of life one from the middle of the third decade and the other embracing the fifth decade of life (2) the lowest fracture loads were most often found in individuals in the eighth and ninth decades of life (3) in human bone lower fracture loads were more frequently obtained in the period of minimal resistance and (4) the average value for fracture strength was found from the 25th through the 35th years of life and corresponds to the average strength from the 50th to the 70th years of life Thus the strength of bone between the 25th and 35th year of life is approximately intermediate to that between the sixth and seventh decades of life

May found when he calculated the coefficient of dispersion that 67 5% of the determinations varied between 0 200 and 0 300

The degree of flexibility of the various specimens was more uniformly distributed than the fracture load. It was particularly interesting that the ulna showed more decreases in flexibility than did the humerus, the femur, and the tibia. The average deflections (micra) for 5 kg. loads were: humerus 226μ, ulna 208μ, femur 222μ, and tibia 213μ.

In general the flexibility shows a rather haphazard increase with the advancing age of the individual from whom the specimen was taken. Maj's studies showed that with the increase of flexibility of the specimens there is a diminution in the amplitude of variability of the fracture loads from the side of major resistance. This inverse relationship was more evident when fracture loads were compared with the elastic deformations of a single skeletal segment.

The flexibility of a bony bar subjected to progressive loading to fracture was not identical in all specimens and three types of moduli of elasticity were demonstrable. In the first type of modulus load and deflection or deformity of the bar until fracture were about equal. Thus, when load was plotted against elastic deformation in millimeters, a straight line was obtained. This type of modulus was theoretical and rather rare.

In the second type of modulus the deformation was approximately proportional to the increase in load up to a certain point, beyond which deformation increased more rapidly than load until fracture occurred. This was the most common type of modulus and gave a concave load-deflection curve.

In the third type of modulus the deformation and load were approximately proportional up to a certain point beyond which load increased more rapidly than deformation. This was a very rare type of modulus and produced a load deflection curve which was convex.

The second type of modulus of elasticity occurred most frequently especially in advanced age. Examples of the first and third types were found in the first half of life and were most prevalent in skeletal segments having a higher fracture strength. From these tests Maj concluded: (1) that the most conspicuous variations in the fracture strength of bones occurred in bony tissue from the same individual rather than between the average re-

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and is a solid structure The porosity of compact bone is also greater than commonly believed as shown by May (1938) who found that specimens of compact bone from metacarpals of an ox had a porosity as high as 30% in some instances Thus referring to compacta as solid bone may be far from true

Fatigue Strength

The first studies ever made on the fatigue strength of bone were those of the author and Milton Lebow who determined the fatigue strength of specimens of compact bone three inches long 0 25 inches wide and 0 090 inches thick At a distance of 0 875 inches from one end of the specimen was an undercut area 0 15 inches in width The dimensions of the specimens were standardized to an accuracy of 0 01 inches The specimens were tested at a stress of 5000 lbs/in² in a Sonntag Flexure Fatigue Testing Machine, model SF 2 An automatic counter on the machine recorded the number of cycles to failure and the machine automatically shut off when the specimen broke During a test the specimen was kept wet by water dripping on it

The specimens were taken from the anterior posterior medial and lateral quadrants of the proximal middle and distal third of amputated unembalmed human femurs, tibias, and fibulas In a preliminary study, presented at the 65th Annual Meeting of the American Association of Anatomists (Evans, *Anat Rec*, *112*, 1952) it was found that at a stress of 5000 lbs/in² the femoral specimens had a life of approximately one million cycles the tibial specimens a life of two million cycles and the fibular specimens a life of three million cycles

In a more extensive study of 46 specimens from the human tibia also tested at a stress of 5000 lbs/in² it was found that the specimens from the middle third of the shaft had the greatest fatigue life (2 015 923 cycles) and those from the proximal third the least (914,749 cycles) With respect to quadrants the specimens from the posterior quadrant were the strongest (2 283 615 cycles) while those from the anterior quadrant were the weakest (1,519 799 cycles) The details of this work will be published later

A study on the fatigue strength of intact human metatarsals

sistance in specimens from individuals of various ages and (2) variations in the mechanical behavior of the bony tissue in individuals of various ages are more than accidental

Maj was unable to give any satisfactory explanation for the rather poor fracture strength of bone from individuals from 25 to 35 years of age He did not believe the differences in density size and structure of Haversian systems nor the prevalent direction of collagen fibers was sufficient to explain the changes occurring in fracture strength of bone with advancing age He believed that the causes of individual variability in fracture strength were primarily biological some acquired, and others congenital He thinks the variability is an explanation of chemical physical diversity situated within the fundamental elements of bony tissue that is the collagen fibers the osteomucoid substance and the mineral salts

The studies of Olivo Maj and Toajari again showed that a long bone is a heterogeneous structure with respect to the bending strength deflections under bending and modulus of elasticity This is true along the length of the diaphysis as well as from the external surface toward the medullary cavity within a single ring cut from the diaphysis Specimens cut longitudinally with respect to the long axis of the intact bone are three times stronger than tangentially cut specimens and six times stronger than radially cut specimens In longitudinally cut specimens the predominate direction of the collagen fibers is parallel with the major axis of the specimen while in radially cut specimens it is not The Italian investigators in some instances reported the deflections occurring in a specimen of compact bone during a bending test in terms of hundredths of a millimeter per 5 kg load In doing this they assumed that deflection was directly proportional to load which may not be true for bone in bending especially when it is wet or has considerable moisture in it

The studies on the bending strength of compact bone provide additional evidence against the results obtained from analyzing stress and strain in bone by means of mathematical analysis of cross sections of bone trajectorial diagrams or photoelasticity because all these methods are based on the assumption that an intact bone is composed of a uniformly homogeneous material

Calabrisi (1949) an average of 149 inch lbs This indicates that embalming reduces the torsion strength of intact bones

The experiments of Olivo, Maj and Toajari with small specimens of compact bone from the metacarpals, metatarsals and femurs of man and other mammals showed there is considerable variation in the bending strength of specimens from different levels of the diaphysis as well as within single rings of bone cut from the whole shaft The bending strength is greatest in rings from the center of the shaft and decreases from there toward the ends of the bone The decrease in bending strength of metatarsals and metacarpals is more gradual from the middle of the shaft toward the proximal end of the bone than it is distal to the center The percentage porosity of the same bones increases from the center of the shaft toward its ends but apparently is not related to the bending strength of the bone The only possible exception may be in those rings of bone in which the porosity is approximately 50%

These studies emphasize the heterogeneous structure of compact bone and the porosity determinations reveal that the compacta is not "solid bone as it is frequently called The relatively low fatigue life of compact bone, as compared with metals is probably related to the presence of Haversian canals and other cavities which make bone far more porous than metals and greatly reduce its fatigue strength

The studies on the torsional shearing and bending strength of bone supplement those on its tensile compressive and direct shearing strength in demonstrating that bone is far from being a homogeneous material Therefore the results on the strength of bone obtained from mathematical analysis of sections of bone from trajectorial diagrams and from photoelastic studies are questionable because all of them are based on the assumption of a solid structure composed of homogeneous material Bone does not have these characteristics

in an attempt to learn something of the mechanism of fatigue fractures is being made by the author and Glenn Lease Fatigue fractures are considered to be the result of repetitive loading such as occurs in walking hence the name march fractures sometimes is applied to them Fatigue fractures have also been reported for the tibia the femur and even the pelvis

Thus far four metatarsals have been tested for fatigue strength *at a load of 10 lbs, two at a load of 12 lbs and 20 at a load of* 15 lbs The latter was found to be the desirable load to use and the following results were obtained from bones tested with a 15 lb load The average life to failure for metatarsals II through V the first being too large to be put in the testing machine was 2 214 833 cycles for the second metatarsal 1 522 600 cycles for the third metatarsal 329,400 cycles for the fourth metatarsal and 215 250 cycles for the fifth metatarsal

These studies are of interest because, as far as is known they are the only ones on the fatigue strength of compact bone considered as a material, and of intact bones However more data are needed before definite conclusions can be drawn

Summary

The average torsional shearing strength of specimens of compact bone of standardized size is considerably less than the average compressive and tensile strength of dry compact bone and the average compressive strength of wet bone However there is some overlap in the range of variation in the torsional shearing strength of dry bone and the tensile strength of wet bone The torsional shearing strength of compact bone is greater than the direct shearing strength of specimens cut perpendicular to the long axis of the bone and loaded parallel with the direction of its fibers However the torsional shearing strength is less than the direct shearing strength of samples cut parallel with the long axis of the bone and loaded perpendicular to the direction of the fibers

The average torque required to fracture fresh femurs by torsion was found by Messerer (1880) to be 1350 inch lbs while for dry embalmed femurs Pedersen Evans and Lissner (1949) reported an average of 348[illegible] inch lbs and Carothers Smith and

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Glossary of Engineering Terms

Biomechanics The science dealing with the effect of force and energy upon organisms or organic material

Couple Two parallel forces equal in magnitude and opposite in direction

Deformation The change in the size or shape of a body produced by the action of a force or absorption of energy

Elastic Limit The point on a stress strain curve up to which a body will return to its original dimensions and shape after removal of a force Beyond this point permanent deformation is produced

Elasticity The ability of a body to return to its original size and shape after the removal of a force

Energy The capacity to do work

Force Anything tending to change the state of rest or motion of a body A push or a pull To completely define a force its point of application its direction and its magnitude must be known

Hysteresis A phenomenon seen in a material subjected to stress which results in less energy being given out by the material in its recovery than was expended on it in producing its deformation The difference is the result of the energy lost as heat A slight temporary set is produced

Mechanics The science dealing with the effect of a force or energy upon the form or motion of a body

Modulus of Elasticity The measurement of the stiffness of a material The units of measurement are lbs /in^2 or kg /cm

Moment of a Force The product of a force and the perpendicular distance to the action line of the force This distance is the lever arm

Proportional Limit The point on a stress strain curve up to which stress and strain are directly proportional

Strain A change in the linear dimensions of a body or material resulting from the application of a force This is called *total strain* The quotient obtained by dividing the total strain by the original length of the body is called *unit strain*

Stress The resistance within a body or material to the deforming effect of a force Stress cannot be seen but its magnitude can be com

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Elasticity. The ability of a body to return to its original size and shape after the removal of a force.

Energy. The capacity to do work.

Force. Anything tending to change the state of rest or motion of a body. A push or a pull. To completely define a force its point of application, its direction, and its magnitude must be known.

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puted in terms of lbs./in.^{2} or kg./cm.^{2} In engineering practice "strength" is commonly used as synonymous with "stress."

Stress-Strain Curve. The curve obtained by plotting stress as the ordinate against strain as the abscissa. The slope of the curve represents the modulus of elasticity of the material.

Torsion. The twisting caused in a body by the application of a couple.

Torque. The twisting moment of a force producing torsion in a body.

Index

Ainsworth W H 133
Allison N 119 120 169 170
Amprino R 127 128
Apfelbach C W 51
Aran, F A 49

Barati A 128
Baker S L 133 134
Bass n A L. 161 162
Becker R B 162
Bell G H 163 169 171 174
Bending in skull 36 38-40
in long bones 56 58 60 62 64 67 71 72
Benninghoff A 51 57 58 97 99 121 123 128
Best C H 182 183
Bethke R M 173
Bone architecture of 111 129
breaking strength intact bones
effect of diet on 160 168
effect of disuse on 169 172
effect of fluorosis on 172 174
effect of healing on 172
effect of hormones on 168 169
of femur 82 91
of pelvis 97 98
of skull 42-43 49 50
compact (compacta)
relation to spongiosa 122
ultimate strength
bending 205 225
compressive 192 197
fatigue 225 226
shearing 199 201
tensile 177 192
torsion 203 204
spongy (spongiosa)
compressive strength 197 199
Bones stress strain in
living bone 14 71
methods of study
colophonium method 11 12
extensometers 14
mathematical analysis 9
photoelasticity 10 11
strain gages 14 15
stresscoat 12 14
Bosworth T J 174
Braus H 96
Brooks B 119 120 169 170
Bruns v V 50 51

Calabrisi P 179 187 188 190 191 194 197 204 227
Carey E J 9 116-117 129
criticism of 117
Carothers C O 179 187 190 191 194 196 204 226
Chambers J W 165 167 174
Charnley J 133 134
Chick limb bud grafts 119 159
Clarke M F 161 162
Colonna P C 140 144
Colophonium technique 11 12
studies of femur by 54 56
studies of tibia by 56
studies of humerus by 56
studies of radius by 56
studies of clavicle by 56
Column behavior under load 6-7
Compression arthrodesis 133 134
Compressive force definition 4
Compressive stress and strain
see stress strain
Contact compression factor 130 133
Coolbaugh C C 14 71
Couple definition 4
magnitude of 5
Culmann analysis of femur 9 10 59 112 114 116 118 128
criticisms of 59 112 114
Curve load-deflection 79 101 102 105
stress strain 7 182 184 186
Cuthbertson D P 163 165 168 169 171 173

Dawson I M 165 166 174

deForest A V 12
Dempster W T 9 184 186-187 190 191 196-197 204
Depression skull fracture 44 45
Disuse effects of 119 121 127 128
Dog experiments on femur 134
experiments on ilium 134 137
experiments on ulna 137 139
immobilization of forelegs 119 120 127 128
skull 34 36 40 42
Dowgjillo N D 99 122
DuBrul E L 101
Dynamic loading of skull 8 34-49
of long bones 56-57 59 61 67

Edgington H H 173 174
Eggers G W N 130 134
Elasticity definition 7
modulus of 7
Ellis G 12
Energy absorption time in skull fracture 41 42
computation of magnitude 8
definition of 7
relation to moisture in bone 182
relation to skull fracture 42 43
relation to strain in bone 182
Engstrom A 127
Evans F Gaynor 9 14 15 60 72 80 83 88 92 94 97 98 100 101 104 105 179 184 186-188 192 200 201 204 216 225-226
Extensometers femoral studies with 14 73 80

Felizet G M 49
Fell H B 148 149
Femur human
architecture of 111 116
breaking strength 82 91
colophonium studies of 54 59 123 125
Culmann s analysis of 9 10 59 112 114 116 118 128
fractures of 82 91
magnitude of strain in 73 81
mathematical analysis of stress in 16-24
modulus of elasticity in 24
photoelastic studies of 28 31
strain distribution in 54 72
stresscoat studies of 60 71
Femur non human
chorio allantoic grafts of 119
stresscoat studies of 68 71
Fibula photoelastic studies of 26-27
tensile strength of 177 183 187
Force definition of 4
magnitude of 4 5
moment of 5
types of 4
Ford L T 134 135
Fracture mechanism in long bones 82 91
in skull 36-53
in pelvis 97 98
Fractures
femoral
abduction 84 86
intertrochanteric 84 86 89 90
neck 83 86
subcapital 84
torsion (spiral) of shaft 87 88
transverse shaft 88
trochanteric 88
pelvic 97 98
skull 34 53
French C 137 144 146
Friedenberg Z B 137 140 146

Gahlco G L 3 111
Gelbke H 144 146
Gillespie J A 121 170 172
Glegg R E 158
Glucksman A 147 150 155 158
Grunewald J B 17 21 23 25
Gurdjian E S 13 15 34 48 50 51 96 98

Hallermann 11 26-29 33
Ham A 128
Hardinge M G 197 199
Harvey B C 160 161
Haversian systems
see osteones
Hayes J F 14 66 80
Henckel K O 122

Hirsch C 102 104 106-110
Howes E L 160 161 172
Hulsen K K [illegible] 24 179 187 189 191
Humerus colophonium studies of 56
  in chorio-allantoic grafts 119
  photoelastic studies of 30-32
Humphry G M 111
Huxley J 119
Hysteresis in intervertebral discs 102

Immobilization effect on bone histology 127
  of dog forelegs 119 120
  of kitten hindlegs 121
  of rat legs 121
Inertia axis of 21
  moment of 21
Intervertebral disc 101 109
  biomechanical behavior of 101 109
  deformations in 101 109
  effect of pressure on 101 105
  hysteresis in 102
  stress in 103 105
Intracranial pressure changes in dogs 41-42

Janssen M 9 115 116 129
Johnson R W 139
Jores L 155

Key J A 134 137
Kick C H 173 174
Koch J C 9 16 25 113 114
Kokott W 155 157
Kuntscher G 11 12 14 25 26 33 54 59 68 73 75 82 83 98 114 123

Lacroix P 157 158
Landauer W 147 149
Lease G 226
Leblond C P 158
Lebow M [illegible] 14 71 80 179 184 186-188 192 200 201 216 225
LeCount E R 51
Leriche R 155
Liddicoat R T 9 184 186-187 190 191 196-197 204
Lindsay M K 160 161 172

Lissner H R 13 15 34-48 50 51 60 65 68 83 86 92 98 104 105 201 226
Loeschcke H 153 157
Lorenz bifurcation 28
Lottes J C 134 137

McKeown R M 160 161
Magnitude of strain
  extensometer measurements 73 80
  stresscoat measurements 80 81
Maj G 206-214 219 225 227
Mandible architecture of 123 125
  colophonium studies of 98
  split line studies of 99 100
  stresscoat studies of 100 101
  trajectories of 123
Marique P 9 14 16-17 19 25 75-80
Mathematical analysis of stress
  criticism of method 21 25
  difficulties in 16
  Grunewald s results 18 19
  Koch s results 16-18
  Marique s results 19 21
Mechanical stimuli in osteogenesis 157 159
Messerer O 49 50 57 82 204 226
Meyer H von 9 112 114 116 122 128 129
Milch H 11 26-29 33
Modulus of elasticity definition of 7
Moment of a force [illegible]
Moment of inertia 20 23
  definition 21
  determination of 21
Murray P D F 99 117 119 121 122 147 150 155 157

Nachemson A 106-110
Neal W M 163
Nucleus pulposus 104

Olivo O M 206 209 210 224 227
Osteones effect of blood supply on 128
  effect of immobilization on 127
  functions of 57 58
  orientation of 122
  relation to split lines 122
  relation to trabeculae 58 59 122

Orr J 163 165 168 171 173
Owen J T 88

Pauwels F 11 26 30 33 96-97 124
 126
Pedersen H E 14 61 65 83 86 98
 204 226
Peirce A W 174
Pelvis 92 98
 fractures of 97 98
 strain in 92 97
 stresscoat studies of 92 97
Perry W L M 168
Phillips P H 174
Photoelasticity 10 11 26 33
 criticism of 33
 studies of femur by 28 31
 studies of fibula by 26-27
 studies of humerus by 31 32
 studies of tibia by 26
Policard A 155
Pomerat C M 130 133
Powers J E 14 66 80
Prediction of fracture site 48 49
Pressure see stress compressive
Principles of mechanics 4 8

Rabbits experiments on bones of 120
Rats experiments on bones of 120 121
 experiments on skulls of 130 132
Rauber A A [illegible] 22 90 177 178 187
 194 196-197 199 200 203 205
Rawling L B 49
Reynolds C F 134
Roux W 9 10 26 29 114 115 128
 129 155
Rowbotham G F 51
Rubber model of a bone
 criticism of use of 26
 use in stress strain analysis 26

Scalp effect on skull fracture 43
Schanz osteotomy 28
Seipel C M 99 122 123
Selby D 119 147 149 150
Shealy A L 162
Shearing force definition 4
Shindler T O 130 133
Sicher H 101
Skull dog
 deformations in 41-42
 intracranial pressure in 41 42
 stresscoat studies on 34 36
Skull human
 breaking strength of 42 43 49 50
 deformation time of 41-42 44
 dynamic loading studies 34 49
 energy absorption of 36-37 42 45
 fracture mechanism 36-53
 quantitative determinations of de
 formations 36 37
 split line studies of 51 53
 static loading studies 49 51
 strain distribution in 34 41
 stresscoat studies of 36-38 46 49
Smith A H 161 162
Smith F C 179 187 188 190 191
 194 196 204 226
Smith L D 88 90
Spears G N 88
Split line technique description of
 121 122
 criticism of mechanical interpretation
 of 51 53 123 126
 studies of femur by 123 125
 studies of mandible by 123
 studies of skull by 51 53
Static loading definition [illegible]
 studies of human femur by 54 56
 60 62 69 70 82 85 88 89
 studies of human pelvis by 97 98
 studies of human skull by 49 51
Strain definition of 5
 distribution of in femur 54 56 58
 distribution of in mandible 98 101
 distribution of in pelvis 92 98
 magnitude of in femur 74 77 80 81
 magnitude of in intervertebral discs
 101 110
 relation to fracture of long bones
 82 91
 relation to fracture of skull 36 39
Strain gage electric
 principle of 15
 skull deformation studies with 41 44
 tibial deformation studies with 71 72
Strain sensitive lacquer 11 14
 long bone studies with 54 71

skull studies with 34 38 46-49 53
Straus W L Jr 68
Stress definition 5
in bone healing 130 140
in bone growth 140 146
in embryonic osteogenesis 147 158
in femur 80 81
in skull fracture 36
levels in human skull 45-47
Stresscoat deformation patterns
in dry and living dog and monkey skulls 34 36
in femur 60 72
in human skull 36-41
in mandible 98 101
in pelvis 93 97
Stresscoat technique 12 13
magnitude of strain 80 81
studies of dog skull by 34 36
studies of human femur by 59 71 80 81
studies of human pelvis by 92 98
studies of human skull by 36-38 46-49 53
studies of monkey skull by 34 36
Stress strain
compressive
in bone growth 140 146 158 159
in bone healing 130 140
in embryonic osteogenesis 147 155
in the femur 54 56
tensile
in bone growth 144 146
in embryonic osteogenesis 147 148 152 155 158 159
in femoral fractures 82 88
in pelvic fractures 97 98
in skull fractures 37 41 46-49
theories of influence in osteogenesis 155 159
Stress strain curve 7
for compact bone 181 182 184 186 202
Strobino L J 140 144
Studitsky A N 147 149

Tappen N C 51 53 122
Taylor N B 162 183
Tensile force definition 4
Tensile strain
dog skull 34 36 41
human skull 37 41 46 49
Thoma R 155 156
Tibia experiments on growth of 140 144 146
Toajan E 206-210 214 219 224 227
Torsion definition 4
fractures 87 88
tests of deformed bones 56 57
tests of femur 56 65 66
tests of tibia 56
Trabeculae
effect of immobilization in 119
in femur 111 116 118
in knee joint 114 115
relation to osteones 57 59
Traction braces effect on stress in bones 30 32
Trajectorial diagrams Culmann s of femur 9 59 112 114 118
criticism of 59 112 114 128 129
Roux s of ankylosed knee 10 26 114 115
Trajectorial theory 9 10 29 59 112 118 123 124 128
criticism of 112 115 117 119 123 129
Triepel H 117 119

Urist M R 139

Vigliani F 127
Virgin W J 101 102
Vertebral column 101 109

Wagstaffe W W 114
Walmsley T 60
Ward F O 111
Webster J E 37-40 42 48 50 51 96 98
Weidenreich F 155 157
Weinnoldt H 155 157
Weir J B de 166-167 172 174
Wermel J 120 121
Wertheim M C 9 177 185 188 192
Wolff J 9 29 112 115 118 128 129
Wolff's Law 115
Wyman J 111

Orr J 163 165 168 171 173
Owen J T 88

Pauwels F 11 26 30-33 96-97 124 126
Pedersen H E 14 61 65 [redacted] 86 98 204 226
Peirce A W 174
Pelvis 92 98
  fractures of 97 98
  strain in 92 97
  stresscoat studies of 92 97
Perry W L M 168
Phillips I H 174
Photoelasticity 10 11 26-33
  criticism of 33
  studies of femur by 28 31
  studies of fibula by 26 27
  studies of humerus by 31 32
  studies of tibia by 26
Policard A 155
Pomerat C M 130 133
Powers J E 14 66 80
Prediction of fracture site 48 49
Pressure see stress compressive
Principles of mechanics 4 8

Rabbits experiments on bones of 120
Rats experiments on bones of 120 121
  experiments on skulls of 130 132
Rauber A A 9 22 90 177 178 187 194 196-197 199 200 203 205
Rawling L B 49
Reynolds C F 134
Roux W [redacted] 10 26 29 114 115 128 129 155
Rowbotham G F 51
Rubber model of a bone
  criticism of use of 26
  use in stress strain analysis 26

Scalp effect on skull fracture 43
Schanz *osteotomy* 28
Seipel C M 99 122 123
Selby D 119 147 149 150
Shealy A L 162
Shearing force definition 4
Shindler T O 130 133
Sicher H 101
Skull dog
  deformations in 41 42
  intracranial pressure in 41 42
  stresscoat studies on 34 36
Skull human
  breaking strength of 42-43 49 50
  deformation time of 41 42 44
  dynamic loading studies 34-49
  energy absorption of 36-37 42-45
  fracture mechanism 36-53
  quantitative determinations of deformations 36 37
  split line studies of 51 53
  static loading studies 49 51
  strain distribution in 34 41
  stresscoat studies of 36-38 46-49
Smith A H 161 162
Smith F C 179 187 188 190 191 194 196 204 226
Smith L D 88 90
Spears G N 88
Split line technique description of 121 122
  criticism of mechanical interpretation of 51 53 123 126
  studies of femur by 123 125
  studies of mandible by 123
  studies of skull by 51 53
Static loading definition 8
  studies of human femur by 54 56 60 62 69 70 82 85 88 89
  studies of human pelvis by 97 98
  studies of human skull by 49 51
Strain definition of 5
  distribution of in femur 54 56 58
  distribution of in mandible 98 101
  distribution of in pelvis 92 93
  magnitude of in femur 74 77 80 81
  magnitude of in intervertebral discs 101 110
  relation to fracture of long bones 82 91
  relation to fracture of skull 36 39
Strain gage electric
  principle of 15
  skull deformation studies with 41 44
  tibial deformation studies with 71 72
Strain sensitive lacquer 11 14
  long bone studies with 54 71

This Book

STRESS and STRAIN in BONES

By

F GAYNOR EVANS PH D

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